WEST Search History

Hide Items | Restore | Clear | Cancel

DATE: Sunday, March 06, 2005

Hide?	Set Name	Query	Hit Count
		DB=PGPB,USPT,USOC,EPAB,JPAB,DWPI,TDBD; PLUR=YES; OP=ADJ	
☐	L21	(5451876 or 4999581 or 4857846)	54
☐	L20	L18 and ((combin$3 or "multi" or coil) with (image or reconstruct$4))	12
☐	L19	L18 and ((combin$3 or (multi with coil)) with (image or reconstruct$4))	9
☐	L18	L17 and ((without or "no" or "off" or "non") with (phase with encod$4))	19
☐	L17	L16 and ((phased or phase) with (array or plurality or multi or multiple or "more than one" or dual or "at least two" or plural) with (receiv$3 or detect$3 or sens$3 or reception or sensitiv$5 or coil))	35
☐	L16	L15 and (gradient)	67
☐	L15	L14 and (phase with encod$4)	73
☐	L14	L13 and ((reference or calibrat$4 or precondition$3 or prescan$4 or prepara$4 or pre-scan$4 or pre-pulse or preliminary or initial$2) with (data or raw or image or k-space or kspace or "k space" or "kx" or "ky" or "kz"))	369
☐	L13	L12 and ((array or plurality or multi or multiple or "more than one" or dual or "at least two" or plural) with (receiv$3 or detect$3 or sens$3 or reception or sensitiv$5 or coil) with (independent$2 or individual$2 or separat$2))	688
☐	L12	L9 and ((correct$4 or compensat$4 or adjust$4 or fix$3 or modify or modif$3 or chang$3 or shift$4 or drift$3 or offset$4 or delay$3 or echo$3) with (amplitude or magnitude or peak))	2225
☐	L11	L9 and ((correct$4 or compensat$4 or adjust$4 or fix$3 or modify or modif$3 or chang$3 or shift$4 or drift$3 or offset$4 or delay$3 or echo$3) with (amplitude or magnitude or peak))	49
☐	L10	L9 and ((correct$4 or compensat$4 or adjust$4 or fix$3 or modify or modif$3 or chang$3 or shift$4 or drift$3 or offset$4 or delay$3) with (amplitude or magnitude or peak))	2139
☐	L9	L8 and (correct$4 or compensat$4 or adjust$4 or fix$3 or modify or modif$3 or chang$3)	5518
☐	L8	L6 and ((receiv$3 or detect$3 or sens$3 or reception or sensitiv$5 or coil) with (independent$2 or individual$2 or separat$2))	5587
☐	L7	L6 and ((array or plurality or multi or multiple or "more than one" or dual or "at least two" or plural) with (receiv$3 or detect$3 or sens$3 or reception or sensitiv$5 or coil) with (independent$2 or individual$2 or separat$2))	1676
☐	L6	L5 and ((array or plurality or multi or multiple or "more than one" or dual or "at least two" or plural) with (receiv$3 or detect$3 or sens$3 or reception or sensitiv$5 or coil))	10142
☐	L5	L4 and (array or plurality or multi or multiple or "more than one" or dual or "at least two")	29433

☐	L4	L3 and (receiv$3 or detect$3 or sens$3 or reception)	39402
☐	L3	L2 and (independent$2 or individual$2 or separat$2)	53114
☐	L2	L1 and (amplitude or magnitude or peak)	60931
☐	L1	((magnetic adj resonance) or MRI or NMR)	198518

END OF SEARCH HISTORY

Hit List

Clear | Generate Collection | Print | Fwd Refs | Bkwd Refs | Generate OACS

Search Results - Record(s) 1 through 19 of 19 returned.

☐ 1. Document ID: US 20050033154 A1

Using default format because multiple data bases are involved.

L18: Entry 1 of 19 File: PGPB Feb 10, 2005

PGPUB-DOCUMENT-NUMBER: 20050033154
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20050033154 A1

TITLE: Methods for measurement of magnetic resonance signal perturbations

PUBLICATION-DATE: February 10, 2005

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
deCharms, Richard Christopher	Montara	CA	US	

US-CL-CURRENT: 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw De

☐ 2. Document ID: US 20040155652 A1

L18: Entry 2 of 19 File: PGPB Aug 12, 2004

PGPUB-DOCUMENT-NUMBER: 20040155652
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20040155652 A1

TITLE: Parallel magnetic resonance imaging techniques using radiofrequency coil arrays

PUBLICATION-DATE: August 12, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Sodickson, Daniel K.	Newton	MA	US	

US-CL-CURRENT: 324/307; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw De

☐ 3. Document ID: US 20040044280 A1

L18: Entry 3 of 19 File: PGPB Mar 4, 2004

PGPUB-DOCUMENT-NUMBER: 20040044280
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20040044280 A1

TITLE: Methods & apparatus for magnetic resonance imaging

PUBLICATION-DATE: March 4, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Paley, Martyn	Keighly		GB	
Lee, Kuan	Sheffield		GB	

US-CL-CURRENT: 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw D

☐ 4. Document ID: US 20030206648 A1

L18: Entry 4 of 19 File: PGPB Nov 6, 2003

PGPUB-DOCUMENT-NUMBER: 20030206648
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20030206648 A1

TITLE: Method and system for image reconstruction

PUBLICATION-DATE: November 6, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
King, Kevin Franklin	New Berlin	WI	US	
Angelos, Elisabeth	Hartland	WI	US	

US-CL-CURRENT: 382/128

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw D

☐ 5. Document ID: US 20020158632 A1

L18: Entry 5 of 19 File: PGPB Oct 31, 2002

PGPUB-DOCUMENT-NUMBER: 20020158632
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020158632 A1

TITLE: Parallel magnetic resonance imaging techniques using radiofrequency coil arrays

PUBLICATION-DATE: October 31, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Sodickson MD Ph.D., Daniel K.	Cambridge	MA	US	

US-CL-CURRENT: 324/307; 324/309, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw D

☐ 6. Document ID: US 20020101236 A1

L18: Entry 6 of 19 File: PGPB Aug 1, 2002

PGPUB-DOCUMENT-NUMBER: 20020101236
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020101236 A1

TITLE: Method for resistivity well logging utilizing nuclear magnetic resonance

PUBLICATION-DATE: August 1, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Wollin, Ernest	Leesburg	FL	US	

US-CL-CURRENT: 324/303

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw D

☐ 7. Document ID: US 20010043068 A1

L18: Entry 7 of 19 File: PGPB Nov 22, 2001

PGPUB-DOCUMENT-NUMBER: 20010043068
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20010043068 A1

TITLE: Method for parallel spatial encoded MRI and apparatus, systems and other methods related thereto

PUBLICATION-DATE: November 22, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Lee, Ray F.	Clifton-Park	NY	US	

US-CL-CURRENT: 324/309; 324/307, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw D

☐ 8. Document ID: US 6841998 B1

L18: Entry 8 of 19 File: USPT Jan 11, 2005

US-PAT-NO: 6841998
DOCUMENT-IDENTIFIER: US 6841998 B1

TITLE: Magnetic resonance imaging method and apparatus employing partial parallel acquisition, wherein each coil produces a complete k-space datasheet

DATE-ISSUED: January 11, 2005

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Griswold; Mark	97318 Kitzingen			DE

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 9. Document ID: US 6771067 B2

L18: Entry 9 of 19 File: USPT Aug 3, 2004

US-PAT-NO: 6771067
DOCUMENT-IDENTIFIER: US 6771067 B2

TITLE: Ghost artifact cancellation using phased array processing

DATE-ISSUED: August 3, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Kellman; Peter	Bethesda	MD		
McVeigh; Elliot	Phoenix	MD		

US-CL-CURRENT: 324/307; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 10. Document ID: US 6717406 B2

L18: Entry 10 of 19 File: USPT Apr 6, 2004

US-PAT-NO: 6717406
DOCUMENT-IDENTIFIER: US 6717406 B2

TITLE: Parallel magnetic resonance imaging techniques using radiofrequency coil arrays

DATE-ISSUED: April 6, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Sodickson; Daniel K.	Newton	MA		

US-CL-CURRENT: 324/307; 324/309, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 11. Document ID: US 6545471 B2

L18: Entry 11 of 19 File: USPT Apr 8, 2003

US-PAT-NO: 6545471
DOCUMENT-IDENTIFIER: US 6545471 B2

TITLE: Method for resistivity well logging utilizing nuclear magnetic resonance

DATE-ISSUED: April 8, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wollin; Ernest	Leesburg	FL		

US-CL-CURRENT: 324/303; 324/300

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 12. Document ID: US 6476606 B2

L18: Entry 12 of 19 File: USPT Nov 5, 2002

US-PAT-NO: 6476606
DOCUMENT-IDENTIFIER: US 6476606 B2

TITLE: Method for parallel spatial encoded MRI and apparatus, systems and other methods related thereto

DATE-ISSUED: November 5, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Lee; Ray F	Clifton-Park	NY		

US-CL-CURRENT: 324/309; 324/307, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 13. Document ID: US 6342784 B1

L18: Entry 13 of 19 File: USPT Jan 29, 2002

US-PAT-NO: 6342784
DOCUMENT-IDENTIFIER: US 6342784 B1

TITLE: Method for resistivity well logging utilizing nuclear magnetic resonance

DATE-ISSUED: January 29, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wollin; Ernest	Leesburg	FL		

US-CL-CURRENT: 324/303; 324/300

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 14. Document ID: US 6166540 A

L18: Entry 14 of 19 File: USPT Dec 26, 2000

US-PAT-NO: 6166540
DOCUMENT-IDENTIFIER: US 6166540 A

TITLE: Method of resistivity well logging utilizing nuclear magnetic resonance

DATE-ISSUED: December 26, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wollin; Ernest	Leesburg	FL		

US-CL-CURRENT: 324/300; 324/303, 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 15. Document ID: US 5929637 A

L18: Entry 15 of 19 File: USPT Jul 27, 1999

US-PAT-NO: 5929637
DOCUMENT-IDENTIFIER: US 5929637 A

TITLE: Flow velocity calculating method in magnetic resonance imaging apparatus

DATE-ISSUED: July 27, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Taguchi; Junichi	Sagamihara			JP
Watanabe; Shigeru	Ibaraki-ken			JP
Sano; Koichi	Yokohama			JP

US-CL-CURRENT: 324/306; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

16. Document ID: US 5451876 A

L18: Entry 16 of 19 File: USPT Sep 19, 1995

US-PAT-NO: 5451876
DOCUMENT-IDENTIFIER: US 5451876 A

TITLE: MRI system with dynamic receiver gain

DATE-ISSUED: September 19, 1995

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Sandford; Lorraine V.	Champaign	FL		
Maier; Joseph K.	Milwaukee	WI		
Stormont; Robert S.	Waukesha	WI		

US-CL-CURRENT: 324/322; 324/314

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

17. Document ID: US 5349296 A

L18: Entry 17 of 19 File: USPT Sep 20, 1994

US-PAT-NO: 5349296
DOCUMENT-IDENTIFIER: US 5349296 A

TITLE: Magnetic resonance scan sequencer

DATE-ISSUED: September 20, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Cikotte; Leonard J.	Solon	OH		
Dannels; Wayne R.	Richmond Heights	OH		
McBride; Thomas R.	Newbury	OH		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

18. Document ID: US 4999581 A

L18: Entry 18 of 19 File: USPT Mar 12, 1991

US-PAT-NO: 4999581
DOCUMENT-IDENTIFIER: US 4999581 A

TITLE: Magnetic resonance imaging system

DATE-ISSUED: March 12, 1991

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Satoh; Kozo	Yokohama			JP

US-CL-CURRENT: 324/309; 324/314

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

19. Document ID: US 4857846 A

L18: Entry 19 of 19 File: USPT Aug 15, 1989

US-PAT-NO: 4857846
DOCUMENT-IDENTIFIER: US 4857846 A

TITLE: Rapid MRI using multiple receivers producing multiply phase-encoded data derived from a single NMR response

DATE-ISSUED: August 15, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Carlson; Joseph W.	San Francisco	CA		

US-CL-CURRENT: 324/309; 324/314

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

Clear | Generate Collection | Print | Fwd Refs | Bkwd Refs | Generate OACS

Term	Documents
WITHOUT	6698843

WITHOUTS	24
NO	12164358
NOES	1082
NOS	705392
NOE	8255
OFF	3356707
OFFS	26252
NON	3972188
NONS	478
PHASE	1848565
(L17 AND ((WITHOUT OR "NO" OR "OFF" OR "NON") WITH (PHASE WITH ENCOD$4))).PGPB,USPT,USOC,EPAB,JPAB,DWPI,TDBD.	19

There are more results than shown above. Click here to view the entire set.

Display Format: - Change Format

Previous Page Next Page Go to Doc#

L18: Entry 16 of 19 File: USPT Sep 19, 1995

US-PAT-NO: 5451876
DOCUMENT-IDENTIFIER: US 5451876 A

TITLE: MRI system with dynamic receiver gain

DATE-ISSUED: September 19, 1995

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Sandford; Lorraine V.	Champaign	FL		
Maier; Joseph K.	Milwaukee	WI		
Stormont; Robert S.	Waukesha	WI		

ASSIGNEE-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY	TYPE CODE
General Electric Company	Milwaukee	WI			02

APPL-NO: 08/ 138273 [PALM]
DATE FILED: October 18, 1993

INT-CL: [06] G01 V 3/00

US-CL-ISSUED: 324/322; 324/314
US-CL-CURRENT: 324/322; 324/314

FIELD-OF-SEARCH: 324/300, 324/307, 324/309, 324/313, 324/314, 324/318, 324/322, 128/653.5

PRIOR-ART-DISCLOSED:

U.S. PATENT DOCUMENTS

Search Selected Search ALL Clear

PAT-NO	ISSUE-DATE	PATENTEE-NAME	US-CL
4700138	October 1987	Shimazaki et al.	324/322
4806866	February 1989	Maier	324/313

OTHER PUBLICATIONS

Effective Dynamic Range Improvement of NMR Signal Detection by Using Analog Programmable Attenuators, C. H. Oh, et al., SMRM Eighth Annual Meeting, (1989), p. 181.

ART-UNIT: 225

PRIMARY-EXAMINER: Arana; Louis M.

ATTY-AGENT-FIRM: Quarles & Brady

ABSTRACT:

An NMR system includes a transceiver which receives NMR signals of varying amplitude during a scan. The gain of the receiver is dynamically changed during the scan to provide an optimal SNR figure without overranging the transceiver's A/D converter. The acquired NMR signals are normalized prior to image reconstruction using correction factors for gain and phase stored in a normalization table.

6 Claims, 5 Drawing figures
Exemplary Claim Number: 1
Number of Drawing Sheets: 3

BRIEF SUMMARY:

1 BACKGROUND OF THE INVENTION

2 The field of the invention is nuclear magnetic resonance imaging methods and systems. More particularly, the invention relates to the adjustment of receiver gain to obtain low-noise images under varying signal conditions.

3 When a substance such as human tissue is subjected to a uniform magnetic field (polarizing field B.sub.0), the individual magnetic moments of the spins in the tissue attempt to align with this polarizing field, but precess about it in random order at their characteristic Larmor frequency. If the substance, or tissue, is subjected to a magnetic field (excitation field B.sub.1) which is in the x-y plane and which is near the Larmor frequency, the net aligned moment, M.sub.2, may be rotated, or "tipped", into the x-y plane to produce a net transverse magnetic moment M.sub.t. A signal is emitted by the excited spins after the excitation signal B.sub.1 is terminated and this signal may be received, digitized and processed to form an image.

4 When utilizing these signals to produce images, magnetic field gradients (G.sub.x G.sub.x and G.sub.z) are employed. Typically, the region to be imaged is scanned by a sequence of measurement cycles in which these gradients vary according to the particular localization method being used. The resulting set of received NMR signals are digitized and processed to reconstruct the image using one of many well known reconstruction techniques.

5 The amplitude of the received NMR signal used to reconstruct an image will vary greatly depending on a number of factors. For example, the NMR signal amplitude will increase with increased slice thickness, increased pulse repetition time (TR), decreased echo time (TE) increased patient size, increased fat content in patient and choice of receiver coil (i.e. whole body coil, surface coil, head coil, etc.). These factors remain relatively constant during the scan and the receiver gain is typically set to a single value during a prescan process, which insures that the peak NMR signal amplitude will not over-range the analog-to-digital converter. Such a prescan is disclosed, for example, in U.S. Pat. No. 4,806,866 entitled "Automatic RF Frequency Adjustment For Magnetic Resonance Scanner".

6 The quality of the reconstructed image is related to its signal-to-noise ratio (SNR), and this SNR may be degraded when the receiver gain is lowered to handle the largest signals. This occurs because the noise figure within the MRI system receiver increases as the receiver gain decreases. During a typical scan the NMR signals from different views will have a range of amplitudes and the receiver gain is fixed at a value which utilizes the full range of the analog-to-digital converter when the maximum expected signal amplitude is received. This means that many low level NMR signals are acquired during the scan without utilizing the full range of the analog-to-digital converter, yet have less than the optimal SNR figure. For example, views with minimal phase encoding gradients applied have high amplitudes and must be acquired with low system gain yielding a less than optimal noise figure. The views acquired with high phase encoding gradients, on the other hand, have small amplitudes, but are acquired with a system noise figure set for the largest signal conditions.

7 SUMMARY OF THE INVENTION

8 The present invention relates to an improved MRI system in which the receiver gain is dynamically adjusted during a scan to optimize the SNR for each received NMR signal. Prior to image reconstruction the NMR signals thus acquired are adjusted to normalize out the differences in amplitude and phase caused by the changing receiver gain settings. More particularly, each NMR signal is acquired during the scan with an associated receiver gain setting that is determined as a function of a scan parameter, each acquired NMR signal is normalized using a value selected from a stored normalization table, and an image is reconstructed from the normalized NMR signals acquired during the scan. One such scan parameter, for example, is phase encoding magnetic field gradient amplitude, and the receiver gain settings increase as a function of increasing phase encoding.

9 A general object of the invention is to increase the SNR of MR images. This is accomplished by more efficiently utilizing the full range of the analog-to-digital converter for acquired NMR signals of vastly different amplitudes. Variations in NMR signal amplitude are related to a scan parameter, and the scan parameter is used to select a receiver gain that will fully utilize the analog-to-digital converter range. For example, the scan parameter may be phase encoding value, or it may be a signal indicative of patient respiration or other body function which causes a predictable change in NMR signal amplitude.

10 A more specific object of the invention is to normalize NMR signals acquired during a scan with different receiver gains. A normalization table stores an amplitude correction and a phase correction value for each possible receiver gain setting. The receiver gain setting associated with each acquired NMR signal is used to select the proper amplitude and phase correction values from this normalization table, and these are applied to normalize the NMR signal.

DRAWING DESCRIPTION:

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 is a block diagram of an MRI system which employs the present invention;

FIG. 2 is an electrical block diagram of the transceiver which forms part of the MRI system of FIG. 1;

FIGS. 3A and 3B are schematic representations of two preferred embodiments of a

normalization table stored in the MRI system of FIG. 1 and employed to practice the present invention; and

FIG. 4 is a flow chart of a program executed by the MRI system of FIG. 1 to practice the present invention.

DETAILED DESCRIPTION:

1 DESCRIPTION OF THE PREFERRED EMBODIMENT

2 Referring first to FIG. 1, there is shown the major components of a preferred MRI system which incorporates the present invention. The operation of the system is controlled from an operator console 100 which includes a keyboard and control panel 102 and a display 104. The console 100 communicates through a link 116 with a separate computer system 107 that enables an operator to control the production and display of images on the screen 104. The computer system 107 includes a number of modules which communicate with each other through a backplane. These include an image processor module 106, a CPU module 108 and a memory module 113, known in the art as a frame buffer for storing image data arrays. The computer system 107 is linked to a disk storage 111 and a tape drive 112 for storage of image data and programs, and it communicates with a separate system control 122 through a high speed serial link 115.

3 The system control 122 includes a set of modules connected together by a backplane. These include a CPU module 119 and a pulse generator module 121 which connects to the operator console 100 through a serial link 125. It is through this link 125 that the system control 122 receives commands from the operator which indicate the scan sequence that is to be performed. The pulse generator module 121 operates the system components to carry out the desired scan sequence. It produces data which indicates the timing, strength and shape of the RF pulses which are to be produced, and the timing of and length of the data acquisition window. The pulse generator module 121 connects to a set of gradient amplifiers 127, to indicate the timing and shape of the gradient pulses to be produced during the scan. The pulse generator module 121 also receives patient data from a physiological acquisition controller 129 that receives signals from a number of different sensors connected to the patient, such as ECG signals from electrodes or respiratory signals from a bellows. And finally, the pulse generator module 121 connects to a scan room interface circuit 133 which receives signals from various sensors associated with the condition of the patient and the magnet system. It is also through the scan room interface circuit 133 that a patient positioning system 134 receives commands to move the patient to the desired position for the scan.

4 The gradient waveforms produced by the pulse generator module 121 are applied to a gradient amplifier system 127 comprised of G.sub.x, G.sub.y and G.sub.z amplifiers. Each gradient amplifier excites a corresponding gradient coil in an assembly generally designated 139 to produce the magnetic field gradients used for position encoding acquired signals. The gradient coil assembly 139 forms part of a magnet assembly 141 which includes a polarizing magnet 140 and a whole-body RF coil 152.

5 A transceiver module 150 in the system control 122 produces pulses which are amplified by an RF amplifier 151 and coupled to the RF coil 152 by a transmit/receive switch 154. The resulting signals radiated by the excited nuclei in the patient may be sensed by the same RF coil 152 and coupled through the transmit/receive switch 154 to a preamplifier 153. The amplified NMR signals are demodulated, filtered, and digitized in the receiver section of the transceiver 150. The transmit/receive switch 154 is controlled by a

signal from the pulse generator module 121 to electrically connect the RF amplifier 151 to the coil 152 during the transmit mode and to connect the preamplifier 153 during the receive mode. The transmit/receive switch 154 also enables a separate RF coil (for example, a head coil or surface coil) to be used in either the transmit or receive mode.

6 The NMR signals picked up by the RF coil 152 are digitized by the transceiver module 150 and transferred to a memory module 160 in the system control 122. When the scan is completed and an entire array of data has been acquired in the memory module 160, an array processor 161 operates to Fourier transform the data into an array of image data. This image data is conveyed through the serial link 115 to the computer system 107 where it is stored in the disk memory 111. In response to commands received from the operator console 100, this image data may be archived on the tape drive 112, or it may be further processed by the image processor 106 and conveyed to the operator console 100 and presented on the display 104.

7 Referring particularly to FIGS. 1 and 2, the transceiver 150 produces the RF excitation field B.sub.1 through power amplifier 151 at a coil 152A and receives the resulting signal induced in a coil 152B. As indicated above, the coils 152A and B may be separate as shown in FIG. 2, or they may be a single wholebody coil as shown in FIG. 1. The base, or carrier, frequency of the RF excitation field is produced under control of a frequency synthesizer 200 which receives a set of digital signals (CF) from the CPU module 119 and pulse generator module 121. These digital signals indicate the frequency and phase of the RF carrier signal produced at an output 201. The commanded RF carrier is applied to a modulator and up converter 202 where its amplitude is modulated in response to a signal R(t) also received from the pulse generator module 121. The signal R(t) defines the envelope of the RF excitation pulse to be produced and is produced in the module 121 by sequentially reading out a series of stored digital values. These stored digital values may, in turn, be changed from the operator console 100 to enable any desired RF pulse envelope to be produced.

8 The magnitude of the RF excitation pulse produced at output 205 is attenuated by an exciter attenuator circuit 206 which receives a digital command, TA, from the backplane 118. The attenuated RF excitation pulses are applied to the power amplifier 151 that drives the RF coil 152A. For a more detailed description of this portion of the transceiver 122, reference is made to U.S. Pat. No. 4,952,877 which is incorporated herein by reference.

9 Referring still to FIG. 1 and 2 the NMR signal produced by the subject is picked up by the receiver coil 152B and applied through the preamplifier 153 to the input of a receiver attenuator 207. The receiver attenuator 207 further amplifies the signal by an amount determined by a digital attenuation signal (RA) received from the backplane 118. It is this attenuation signal (RA) that is employed to dynamically adjust the receiver gain during a scan in accordance with the preferred embodiment of the invention.

10 The received signal is at or around the Larmor frequency, and this high frequency signal is down converted in a two step process by a down converter 208 which first mixes the NMR signal with the carrier signal on line 201 and then mixes the resulting difference signal with the 2.5 MHz reference signal on line 204. The down converted NMR signal is applied to the input of an analog-to-digital (A/D) converter 209 which samples and digitizes the analog signal and applies it to a digital detector and signal processor 210 which produces 16bit in-phase (I) values and 16-bit quadrature (Q) values corresponding to the received signal. The resulting stream of digitized I and Q values of the received signal are output through backplane 118 to the memory

module 160 where they are normalized in accordance with the present invention and then employed to reconstruct an image.

11 The 2.5 MHz reference signal as well as the 250 kHz sampling signal and the 5, 10 and 60 MHz reference signals are produced by a reference frequency generator 203 from a common 20 MHz master clock signal. For a more detailed description of the receiver, reference is made to U.S. Pat. No. 4,992,736 which is incorporated herein by reference.

12 The present invention is implemented by changing the digital attenuation signal (RA) applied to the receiver during the scan so that NMR signals of widely varying amplitude can be acquired at an improved SNR. After each NMR signal is acquired, therefore, its measured amplitude must be adjusted to account for the particular receive attenuation (RA) used during its acquisition. This adjustment, or "normalization" of the NMR signal amplitudes insures that the relative amplitudes of the NMR signals employed to reconstruct an image are maintained and they each contribute accurately to the reconstructed image. As will be described below, the amplitude adjustments are made by multiplying the acquired signal by a factor (A) which normalizes it with an NMR signal acquired at the optimal receiver attenuation value (RA).

13 Because the NMR signals are Fourier transformed during the image reconstruction process, the relative phase of the acquired signals must also be maintained. While it is possible to construct a signal attenuator with little or no variation in phase shift between settings, as a practical matter this is not desirable. Thus, practical receivers not only change the amplitude of the acquired NMR signal as a function of the attenuation value (RA), but they also change the time delay imposed on the signal. These time delays must be normalized to preserve the relative phases of all the NMR signals employed to reconstruct an image. Otherwise, the image resolution is reduced due to smearing, or blurring, caused by misplacement of spin signals along the readout gradient axis.

14 Referring particularly to FIGS. 1, 3A and 3B, as each NMR signal is acquired by the transceiver 150 it is stored as an array of complex numbers in the memory module 160. Each of these complex numbers indicates the phase and amplitude of a time domain sample of the NMR signal. As will be described in more detail below, the CPU module 119 operates on this acquired NMR signal to normalize its amplitude and phase in accordance with respective correction factors (A) and (.theta.) stored in a normalization table 225 contained in memory module 160.

15 One preferred embodiment of this normalization table 225 is shown in FIG. 3A and stores time domain corrective values A and .theta. for each possible receive attenuation value (RA). The attenuation value RA at which the stored NMR signal was acquired is employed as an index into this table 225, and the corrective factors A and .theta. are read out and employed to correct each time domain NMR signal sample (I.sub.t, Q.sub.t) as follows:

I.sub.n =I.sub.t (A)cos .theta.-Q.sub.t (A)sin .theta.

Q.sub.n =I.sub.t (A)sin .theta.+Q.sub.t (A)cos .theta. (1)

16 The resulting normalized NMR signal (I.sub.n, Q.sub.n) is then transferred to the array processor 161 which performs the Fourier transformations necessary to reconstruct an image.

17 A second preferred embodiment of the normalization table 225 is shown in FIG. 3B and stores frequency domain corrective values A and .theta. for each possible receive attenuation value (RA) and at each discrete frequency of a Fourier transformed NMR signal. In this embodiment of the invention the acquired NMR signal is first Fourier transformed to the frequency domain by the array processor 161 and stored in memory 160 as an array of complex values in discrete frequency "bins". The attenuation value RA at which the stored NMR signal was acquired is employed as one index into the normalization table 225 of FIG. 3B, and the frequency bin number (f) of a particular NMR signal value is used as a second index to read out the proper correction factors A and .theta.. The correction factors A and .theta. are applied to alter the amplitude and phase of the frequency domain NMR signal samples (I.sub.f, Q.sub.f) to produce normalized samples (I.sub.n, Q.sub.n) as indicated above in equation (1). After all the signal values have been separately corrected, the normalized frequency domain NMR signal is conveyed to the array processor 161 to complete the image reconstruction process. This second embodiment of the invention is preferred when the phase or the amplitude changes imposed on the NMR signal by the receiver are not only dependent on the receive attenuation RA, but also are significantly frequency dependent. In other words, when the phase correction .theta. is not relatively constant for each receive attenuation setting (RA), or amplitude the correction (A) is frequency dependant, the more complex method is preferred.

18 The corrective values A and .theta. in the normalization table 225 are determined for each receiver as part of a calibration process and remain fixed. The normalization table 225 of FIG. 3A is produced by applying a sine wave of constant amplitude A.sub.0 and sampling this signal at each possible receive attenuation setting RA. The corrective values for the normalization table 225 of FIG. 3B are produced in a similar manner, but for each attenuation setting (RA) the frequency of the applied sine wave also is swept through the entire set of readout gradient axis frequencies. In either case the received signal is Fourier transformed to the frequency domain and the complex value (I.sub.f, Q.sub.f) in the frequency bin corresponding to the frequency of the applied sine wave is used to calculate the corrective values for that RA setting and frequency bin as follows: ##EQU1## where A.sub.nom and .theta..sub.nom are the amplitude and phase of the values produced in the same frequency bin by the applied signal measured with RA set to its optimal value from a receiver noise standpoint. The corrective values determined by this receiver calibration process are stored as normalization table 225 in memory module 160 and are used during any subsequent scan in which receiver attenuation (RA) is dynamically changed.

19 It should be apparent to those skilled in the art that when a plurality of receivers are used in parallel, as with a phase array receive coil, a separate normalization table 225 may be created for each and used separately in the subsequent scans.

20 Referring particularly to FIG. 4, the CPU module 119 directs the data acquisition process during a scan in accordance with a stored program. As indicated by process block 230, a prescan is performed first in which NMR data is acquired from which the transceiver is calibrated as is well known in the art. As will be discussed in more detail below, NMR data may also be acquired at this time to determine how NMR signal amplitude will vary during the scan as a function of certain scan parameters such as phase encoding value, slice select location and echo time TE. This information is used to build a table of RA settings which are output to the transceiver 150 as the scan is "played out" and these scan parameters are changed.

21 After the prescan 230, a loop is entered in which scan parameters are output to the pulse generator 121 at process block 232. The appropriate receive attenuation value RA is output to transceiver 150 at process block 234, and the programmed pulse sequence is then initiated as indicated at process block 236 to acquire an NMR signal which is stored in memory module 160 as described above. As indicated at process block 238, this acquired NMR signal is then normalized using one of the above-described procedures and the data is passed to the array processor 161 for image reconstruction. This process repeats after changing the scan parameters at process block 240 until all the NMR signals required by the scan protocol have been acquired as determined at decision block 242. The completed scan is indicated and the image, or images are reconstructed and made available to the operator for viewing or further processing.

22 The table of receiver attenuation values (RA) used during the scan can be produced in a number of ways. Such a table may be constructed, for example, by executing the pulse sequence with no phase encoding, and setting the receive attenuation (RA.sub.0) for an optimal signal level (A.sub.0). The pulse sequence is then repeated with phase encoding applied to measure the NMR signal level (A). The value of receive attenuation (RA) required to produce the optimal signal level (A.sub.0) is then calculated as follows:

RA=RA.sub.0 (A/A.sub.0) (4)

23 This may be repeated for each phase encoding value used during the scan, or preferably, only a few values are measured and RA settings for all possible phase encodings are determined by interpolating between the calculated RA settings. Regardless of the precise method used, the resulting table of RAvalues is stored in the memory module 160 and is used to set the receiver attenuation during the subsequent scan.

24 A similar method may be employed to calculate other receive attenuation tables for use during the scan. For example, if multiple slices are acquired during the scan over a range of human anatomy that produces vastly different signal levels, the signal level is sampled from each slice during the prescan. From these measurements a receive attenuation table is produced which may be used during the subsequent scan to dynamically adjust receiver gain.

25 It should be apparent to those skilled in the art that the receive attenuation can be adjusted dynamically during the scan as a function of more than one scan parameter. This is accomplished by multiplying the receive attenuation values from corresponding receive attenuation tables, and applying the combined values to the transceiver 150 during the scan. This combined value is then used, of course, during the subsequent normalization process.

CLAIMS:

We claim:

1. In an NMR system a method for reconstructing an image from a plurality of acquired NMR signals, the steps comprising:

a) storing a normalization table comprised of a plurality of amplitude and phase corrections (A,.theta.), each of the amplitude and phase corrections being associated with one of a corresponding plurality of receiver attenuation values (RA);

b) acquiring one of said NMR signals with a receiver whose gain is set by one of said receiver attenuation values (RA);

c) normalizing the acquired NMR signal by altering its amplitude and phase by an amount determined by the amplitude and phase corrections (A,.theta.) in the normalization table associated with said one of said receiver attenuation values (RA);

d) repeating steps b) and c) to acquire further NMR signals with the receiver gain set by different ones of said receiver attenuation values (RA); and

e) reconstructing an image using the normalized, acquired NMR signals.

2. The method as recited in claim 1 in which the acquired NMR signal is normalized in step c) by multiplying its amplitude by the amplitude correction (A) and shifting its phase by an amount determined by the phase correction (.theta.).

3. The method as recited in claim 1 in which each NMR signal acquired in step b) is Fourier transformed prior to its normalization in step c).

4. The method as recited in claim 1 in which the different ones of said receiver attenuation values (RA) are selected as a function of a parameter which changes value during the acquisition of said plurality of NMR signals.

5. The method as recited in claim 4 in which the parameter is the value of a phase encoding gradient field which is produced by the NMR system.

6. The method as recited in claim 4 in which the parameter is the location of a slice from which the NMR signal is produced.

L18: Entry 18 of 19 File: USPT Mar 12, 1991

US-PAT-NO: 4999581
DOCUMENT-IDENTIFIER: US 4999581 A

TITLE: Magnetic resonance imaging system

DATE-ISSUED: March 12, 1991

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Satoh; Kozo	Yokohama			JP

ASSIGNEE-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY	TYPE CODE
Kabushiki Kaisha Toshiba	Kawasaki			JP	03

APPL-NO: 07/ 362443 [PALM]
DATE FILED: May 30, 1989

FOREIGN-APPL-PRIORITY-DATA:

COUNTRY	APPL-NO	APPL-DATE
JP	62-244355	September 30, 1987

PCT-DATA:

APPL-NO	DATE-FILED	PUB-NO	PUB-DATE	371-DATE	102(E)-DATE
PCT/JP88/01005	September 30, 1988	WO89/02723	Apr 6, 1989	May 30, 1989	May 30, 1989

INT-CL: [05] G01R 33/20

US-CL-ISSUED: 324/309; 324/314
US-CL-CURRENT: 324/309; 324/314

FIELD-OF-SEARCH: 324/300, 324/307, 324/309, 324/310, 324/311, 324/312, 324/313, 324/318, 324/322

PRIOR-ART-DISCLOSED:

U.S. PATENT DOCUMENTS

Search Selected Search ALL Clear

PAT-NO	ISSUE-DATE	PATENTEE-NAME	US-CL
4620153	October 1986	Hino	324/309

☐	4649346	March 1987	Yeung et al.	324/309
☐	4691162	September 1987	Van Uijen	324/309
☐	4724388	February 1988	Sano	324/309
☐	4736160	April 1988	Sano et al.	324/307
☐	4777625	October 1988	Sakui et al.	365/189
☐	4819207	April 1989	Sakui et al.	365/222

FOREIGN PATENT DOCUMENTS

FOREIGN-PAT-NO	PUBN-DATE	COUNTRY	US-CL
55-112556	August 1980	JP	
60-179643	September 1985	JP	
60-238973	November 1985	JP	
60-263842	December 1985	JP	
61-48752	March 1986	JP	

OTHER PUBLICATIONS

IEEE Journal of Solid-State Circuits, vol. SC-21, No. 5, pp. 618-626, "A 4-Mbit DRAM with Trench-Transistor Cell"; Pallab K. Chatterjee et al; Oct. 1986.

ART-UNIT: 265

PRIMARY-EXAMINER: Tokar; Michael J.

ATTY-AGENT-FIRM: Oblon, Spivak, McClelland, Maier & Neustadt

ABSTRACT:

A magnetic resonance imaging system includes magnetic resonance signal detecting means (1, 2, 3, 4, 7, 8, 10) for detecting a magnetic resonance signal from an object to be examined, receiving means (9) for phase-sensitive detecting and amplifying the magnetic resonance signal, data acquiring means (11) for sampling and digitizing the magnetic resonance signal obtained by the receiving unit (9), and image reconstructing means (12) for performing image reconstruction on the basis of the magnetic resonance signal and the sampling data obtained by the data acquiring means (11). The magnetic resonance imaging system includes a reference signal phase correcting circuit (26) for correcting a phase of a phase-sensitive detecting reference signal, a base line correcting circuit (27) for correcting a base line of the magnetic resonance signal detected by the magnetic resonance signal detecting means (1, 2, 3, 4, 7, 8, 10), and a sampling point correcting circuit (28) for correcting a sampling point of the magnetic resonance signal in the data acquiring means (11).

14 Claims, 11 Drawing figures
Exemplary Claim Number: 1
Number of Drawing Sheets: 8

BRIEF SUMMARY:

1 BACKGROUND OF THE INVENTION

2 1. Field of the Invention

3 The present invention relates to a magnetic resonance imaging system and, more particularly, to a magnetic resonance imaging system in which correction is performed for acquired magnetic resonance data by hardware, thereby reducing a load on software processing except for Fourier transform processing for image reconstruction and improving throughput of the system.

4 DISCUSSION OF BACKGROUND

5 As is well known, magnetic resonance imaging is a technique capable of obtaining chemical and physical microscopic information about molecules by utilizing a phenomenon in which when atomic nuclei having a specific spin and a magnetic moment based on the spin are placed in a uniform static magnetic field, the atomic nuclei resonantly absorb an energy of an RF magnetic field rotating at a predetermined frequency in a plane perpendicular to the direction of the static magnetic field.

6 As a method of visualizing a spatial distribution of specific atomic nuclei (e.g., hydrogen atomic nuclei contained in water and fat) in an object to be examined by using the magnetic resonance imaging, a projecting reconstruction method by Lauterbur, a Fourier method by Kumar, Welti, or Ernst, a spin warp method as a modification of the Fourier method by Hutchison et al., an echo planar method by Mansfield, and the like have been proposed.

7 When the magnetic resonance imaging is to be performed on the basis of these methods, in order to obtain a reconstructed image with high precision and high image quality from acquired magnetic resonance data, correction is performed for various error factors.

8 For example, correction of a base line of a magnetic resonance signal (echo), correction of a sampling point of magnetic resonance data, and phase correction of a detection reference signal can be considered as the correction performed upon acquisition of magnetic resonance data and reconstruction of a magnetic resonance image. As a method of performing these correction operations, a method of performing base line correction, shifting of a sampling point based on an interpolation operation, and phase correction can be performed by software processing for the acquired magnetic resonance data, as pre-processing prior to image reconstruction processing. By Fourier-transforming the corrected magnetic resonance data by the above correction processing, image reconstruction can be properly performed.

9 Especially in a so-called half encoding method in which data of a half of a Fourier space is obtained on the basis of measurement data of the other half by utilizing the fact that data point-symmetrical about the origin of the Fourier space are complex conjugate with each other, a magnetic resonance data error greatly adversely affects a reconstructed image. Therefore, the above correction processing is essential.

10 FIG. 1 shows a procedure of magnetic resonance data acquisition and image reconstruction including the above correction processing based on the half encoding method.

11 First, magnetic echo data (to be referred to as "zero-encoded MR data"

hereinafter) which are not phase-encoded are acquired from an object to be examined or a proper phantom by a magnetic, resonance excitation sequence excluding application of a gradient magnetic field for phase encoding (step S1). On the basis of the acquired zero-encoded MR data, an offset value of an echo signal is obtained, and the offset is corrected (step S2). On the basis of the acquired zero-encoded MR data, a peak point position of a norm of the echo signal is detected (step S3). On the basis of the detected peak point position, a shift amount .DELTA.ts for correcting a sampling point position is obtained (step S4). On the basis of the zero-encoded MR data at the peak point position, a correction phase angle .DELTA..phi.c for a detection reference signal is obtained (step S6). After the above preliminary measurements and correction value determination processing are performed, main measurements and image reconstruction processing are performed as follows.

12 By an imaging magnetic resonance excitation sequence (including application of a phase encoding gradient magnetic field), magnetic resonance data (to be referred to as "kth-encoded MR data" hereinafter) is acquired by kth (k=1, 2, . . .) encoding (step S6). Base line correcting processing and sampling point shift processing based on an interpolation operation and using a sampling point correction amount .DELTA..phi.c obtained in step S4 are performed for the acquired kth-encoded MR data (step S7). Detection reference signal phase correction processing is performed for the data obtained in step S7 by using the correction phase angle .DELTA..phi.c obtained in step S5 (step S8). Steps S6 to S8 are repeatedly performed a predetermined number of times to acquire and correct magnetic resonance data required for image reconstruction based on the half encoding method. On the basis of the magnetic resonance data acquired and corrected as described above, point-symmetrical data in the Fourier space are generated (step S9). The magnetic resonance data obtained in the above processing are two-dimensionally Fourier-transformed to obtain a reconstructed image (step S10).

13 The base line correction and sampling point shifting in step S7 and the detection reference signal phase correction in step S8 correspond to the correction processing described above. These processing operations are performed by software.

14 When the above correction processing is to be performed by software as described above, a long time period is required for calculations and data transfer for executing the above correction processing. Therefore, the processing time must be reduced in order to improve throughput of the magnetic resonance imaging system.

15 Especially in multi-slice imaging for obtaining magnetic resonance data of a plurality of slices of an object to be examined and obtaining a magnetic resonance image of the plurality of slices or in three-dimensional imaging for acquiring magnetic resonance data of a three-dimensional imaging region corresponding to a plurality of adjacent slices of an object to be examined and obtaining magnetic resonance image information of the three-dimensional imaging region, values of a sampling point shift amount .DELTA.ts and a phase angle .DELTA..phi.c for correction differ for the respective slices of the object to be examined. In addition, in the multi-slice and three-dimensional imaging, an amount of data to be processed is enormous for imaging of each slice. Therefore, a strong demand has arisen for reducing a time required for the above correction processing in the multi-slice and three-dimensional imaging.

16 In order to perform the above interpolation operation with sufficient precision, sampling must be performed at a relatively high speed compared to a sampling pitch determined by a Nyquist frequency. For example, in order to

form an image of a (256.times.256) matrix, signal sampling is performed for 1,024 or 512 points. The above various correction processing operations are performed for the sampled data, and the data are selected every four points to obtain 256 points. A two-dimensional Fourier transform operation is then performed to obtain an image. In this case, a memory must have a capacity larger than that originally required. In addition, a time required for the correction processing is prolonged.

17 A magnetic resonance signal changes at high rate near the origin (i.e., near a peak echo) of the Fourier space. For this reason, the magnetic resonance signal near the peak echo must be interpolated with sufficiently high precision. Therefore, high-order interpolation is performed as the interpolation operation. Such high-order interpolation, however, requires a very long calculation time.

18 In addition, the interpolation operation originally includes an error to a certain degree. Therefore, it is desired to obtain correct data in the Fourier space without the interpolation operation, if possible.

19 In a so-called full encoding method in which encoding is performed for the entire Fourier space, an absolute-value image can be obtained from real- and imaginary-part images obtained by Fourier transform without performing an interpolation operation. In this case, not only a calculation time is required for obtaining an absolute value, but also a negative signal region of an inversion recovery image (IR image) is inverted because phase information is lost. For this reason, decisive problems, in which a correct IR image cannot be obtained and a flow rate distribution cannot be imaged using phase information, arise. That is, even when the full encoding method is applied, it is generally required to perform the interpolation operation to obtain a correct real-part image. In this case, substantially the same problems as in the half encoding method in which the interpolation operation is essential to generate data of a half space arise.

20 A computer for use in the magnetic resonance imaging system is normally a so-called mini-computer. Even if an exclusive calculating apparatus is added to such a computer, a calculation time of several seconds is required for image reconstruction of a (256.times.256) matrix. For this reason, the system requiring the above various software correction operations cannot perform image reconstruction within a short time period (e.g., about 10 msec) to display an image of a dynamic object portion such as a heart on a display unit in real time.

21 In recent years, however, since an S/N ratio in magnetic resonance measurement is improved as magnetic resonance measuring techniques have progressed, a flip angle caused by excitation can be set much smaller (e.g., 10.degree.) than 90.degree. which is generally used to sufficiently increase a speed of recovery to a thermal equilibrium state as compared with a spin lattice relaxation (longitudinal relaxation) time (T1). Therefore, unlike in a conventional system, a wait time for recovery to the thermal equilibrium state after magnetic resonance data acquisition can be set to be substantially zero, thereby performing semi-continuous excitation of a spin system.

22 By combining the use of such a small flip angle and a ultra-high-speed imaging method such as the echo planar method and the high-speed Fourier method, it is assumed to be possible to realize a real-time magnetic resonance imaging system. To realize such a system, an image reconstruction time is also desired to be significantly reduced. Therefore, since image reconstruction in the magnetic resonance imaging system is originally only a Fourier transform

operation, it is assumed to be very effective to achieve a system arrangement not requiring pre- or post-processing except for the Fourier transform operation in order to reduce the image reconstruction time.

23 As described above, in the magnetic resonance imaging system, when various correction processing operations are performed by software in a computer, a long time period and an unnecessary memory capacity are required for imaging Therefore, a demand has arisen for a countermeasure against this problem.

24 The present invention has been made to solve the above problem and to provide a magnetic resonance imaging system in which simple hardware is added for correction processing so that the correction processing can be performed by a simple procedure using the hardware to obtain proper magnetic resonance data, and image reconstruction can be performed by merely Fourier transform (two-dimensional Fourier transform in the case of a two-dimensional image, and three-dimensional transform in the case of a three-dimensional image), thereby significantly reducing a time required for the reconstruction.

25 SUMMARY OF THE INVENTION

26 In order to achieve the above object of the present invention, a magnetic resonance imaging system of the present invention comprises a base line correcting circuit for correcting a base line of a magnetic resonance signal, a sampling point correcting circuit for correcting a shift of a sampling point of the magnetic resonance signal (or an application time adjusting circuit for a reading gradient magnetic field), and a reference signal phase correcting circuit for correcting a phase of a phase-sensitive detection reference signal of the magnetic resonance signal. In the magnetic resonance imaging system of the present invention, on the basis of data obtained by a predetermined procedure in a correction data determination mode, a computer supplies a correction value to the above circuits to perform an operation in an imaging mode, thereby automatically obtaining sampling data of a magnetic resonance signal for correct image reconstruction.

27 According to the magnetic resonance imaging system of the present invention, correct image reconstruction magnetic resonance data can be obtained by hardware, e.g., the above three correcting circuits. Therefore, since a magnetic resonance reconstructed image can be obtained by only Fourier transform without performing correction processing using software, a reconstruction time can be significantly reduced. As a result, a diagnostic efficiency is improved, and a more practical real-time magnetic resonance imaging system is realized.

DRAWING DESCRIPTION:

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 is a flow chart for explaining various correction procedures by software processing in a magnetic resonance imaging system; FIG. 2 is a block diagram showing an overall magnetic resonance imaging system; FIG. 3 is a block diagram showing an arrangement of a main part of the magnetic resonance imaging system according to an embodiment of the present invention; FIG. 4 is a flow chart for explaining a correction processing procedure in the embodiment shown in FIG. 3; FIG. 5 is a block diagram for explaining an arrangement of a reference signal phase correcting circuit in the embodiment shown in FIG. 3; FIG. 6 is a block diagram for explaining an arrangement of a base line correcting circuit in the embodiment shown in FIG. 3; FIG. 7 is a block diagram showing an arrangement of a sampling point correcting circuit in the embodiment shown in FIG. 3; FIG. 8 is a timing chart

showing a pulse sequence according to a two-dimensional Fourier method; FIGS. 9(a) and 9(b) are graphs for explaining a data constructing method in a half space based on a point symmetry in accordance with a Fourier method and an echo planar method, respectively; and FIG. 10 is a timing chart showing a pulse sequence of the echo planar method based on a .theta..degree. pulse.

DETAILED DESCRIPTION:

1 DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT

2 An embodiment of the present invention will be described in detail below with reference to the accompanying drawings.

3 FIG. 2 is a block diagram showing an arrangement of a magnetic resonance imaging system according to an embodiment of the present invention.

4 A static magnetic field magnet 1 comprises a plurality of coil units. The static magnetic field magnet 1 is driven by an excitation power source 2 controlled by a system controller 10 and applies a uniform static magnetic field to an object to be examined 5 (e.g., a patient) laid on a bed 6.

5 A gradient magnetic field coil 2 comprises, e.g., a plurality of sets of coil elements The gradient magnetic field coil 2 is driven by a driver 4 controlled by the system controller 10 and applies gradient magnetic fields Gx, Gy, and Gz, whose magnetic field intensities linearly change in orthogonal x and y directions in a desired slice plane and a z direction perpendicular to the slice plane, respectively, to the object to be examined 5 (so as to superpose them on the static magnetic field).

6 Under the control of the controller 10, an RF magnetic field generated from a probe 7 by an RF signal from a transmitting unit 8 is applied to the object to be examined 5. A magnetic resonance signal received by the probe 7 is amplified and detected by a receiving unit 9 and then supplied to a data acquiring unit 11 under the control of the controller 10.

7 That is, as shown in FIG. 3, in the receiving unit 9, the magnetic resonance signal received by the probe 7 is amplified by an RF amplifier 20, quadrature-detected and converted into a video-band signal by an quadrature detector 21, amplified by a video amplifier 22, and supplied to the data acquiring unit 11 via a low-pass filter 23. In the data acquiring unit 11, the magnetic resonance signal supplied from the receiving unit 9 is sampled and digitized by an A/D converter 24. The digitized signal is then supplied to a computer 12 via an interface 25.

8 The computer 12 is operated by an operator via a console 13. On the basis of the sampled magnetic resonance data for image formation supplied from the data acquiring unit 11, the computer 12 performs image reconstruction processing and obtains magnetic resonance image data. The computer 12 controls the system controller 10. The image data obtained by the computer 12 is supplied to an image display 14 and displayed as an image.

9 FIG. 3 shows in detail a part (correcting circuit) related to the present invention in the system controller 10 shown in FIG. 2 and arrangements of the receiving unit 9 and the data acquiring unit 11.

10 The system controller 10 comprises a reference signal phase correcting circuit

26 for correcting a phase of a reference signal to be supplied to the quadrature detector 21, a base line correcting circuit 27 for correcting an offset of a base line of a magnetic resonance signal, and a sampling point correcting circuit 28 for delay-controlling a timing of a sampling clock to be supplied to the A/D converter 24.

11 FIG. 4 shows a procedure for performing imaging on the bases of a pulse sequence of a two-dimensional Fourier method shown in FIG. 8.

12 Prior to an imaging mode operation for final imaging, a correction data determining mode operation for determining correction data is performed.

13 In accordance with a sequence obtained by excluding an encoding gradient magnetic field Ge from the sequence shown in FIG. 8, i.e., in accordance with a zero encoding excitation sequence, an RF pulse rf, a slicing gradient magnetic field Gs, and a reading gradient magnetic field Gr are applied to the object to be examined 5 or a proper phantom prepared for adjustment, and a magnetic resonance echo signal in a zero encoding state is observed, thereby acquiring zero-encoded MR data (step 31). A timing (t1, t2, and t3) for applying the reading gradient magnetic field Gr is set in advance upon adjustment of the pulse sequence so that an echo peak appears at a predetermined timing (t=2.tau.). Similarly, a base line and a phase (an absolute phase and a 90.degree. relative phase in quadrature detection) of a reference signal for detection are properly set upon adjustment. A series of operations from steps 31 to 38 aims at correcting these offsets derived from some causes.

14 By performing data acquisition in a time region in which the echo signal is completely attenuated, an offset value, a base line value Vb(t), is obtained for the zero-encoded MR data (step 32). In order to obtain the base line value Vb(t) with a time variation, data acquisition may be performed at a predetermined timing in a no signal state, e.g., a state in which the gradient magnetic fields Gs and Gr are applied by a sequence not applying the RF pulse rf (when the base line value Vb(t) also varies by phase encoding, a sequence including the encoding gradient magnetic field can be used to obtain the base line value Vb(t) upon encoding). When an S/N ratio (signal-to-noise ratio) is not sufficient upon acquisition of the base line values, averaging is performed by a plurality of times of data acquisition. An offset of the video amplifier 22 shown in FIG. 2 is adjusted by using the determined base line value Vb(t), thereby correcting the base line (step 33). An echo peak is detected by an interpolation operation on the basis of a norm (data corresponding to the size of an echo amplitude vector, i.e., an absolute value of complex data) of the zero-encoded MR data (step 34). An offset of the echo peak from a nearest sampling timing is obtained as .DELTA.ts. A sampling timing of the echo data in the A/D converter 24 is shifted back and forth by .DELTA.ts (step 35), and the zero-encoded MR data are acquired again (step 36). Since the data is already subjected to base line correction and sampling point shift correction by hardware, a correcting phase angle .DELTA..phi.c with respect to a phase error is immediately determined at the echo peak (tr=0) by using a known algorithm, i.e., by tan.DELTA..phi.c=Vi/Vr from complex data Vr+iVi at the echo peak point (step 37). A phase of the quadrature detecting reference signal to be supplied to the quadrature detector 21 shown in FIG. 2 is shifted by .DELTA..phi.c (step 38).

15 By the above procedure, the correction operations of the correcting circuits 26, 27, and 28 shown in FIG. 3 are completed. Therefore, an imaging mode operation will be performed next.

16 In the imaging mode, on the basis of the pulse sequence shown in FIG. 8, phase encoding is performed by Ge in accordance with predetermined steps, thereby acquiring echo data in a half space, e.g., a region of te.gtoreq.0 in the Fourier space shown in FIG. 9(a). Since echo data point-symmetrical about the origin are complex conjugate with each other, echo data in a region of the other half space (te<0) are immediately generated on the basis of the acquired echo data. Thereafter, an desired image can be obtained by only a two-dimensional Fourier transform operation (step 39).

17 Detailed arrangements of the correcting circuits 26, 27, and 28 shown in FIG. 3 will be described below with reference to FIGS. 5, 6, and 7, respectively.

18 FIG. 6 shows the base line correcting circuit 27 and its peripheral circuits. The video amplifier 22 comprises two differential amplifiers 51 and 52. The video amplifier 22 voltage-amplifies a detection output of real and imaginary parts of an echo signal supplied from the quadrature detector 21 and supplies it to the low-pass filter 23. The base line correcting circuit 27 applies the base line value Vb(t) obtained by the computer 12 in step 32 shown in FIG. 4 to the inverting input terminals of differential amplifiers 51 and 52 via an interface 51 and D/A converters 54 and 55, thereby automatically performing base line correction.

19 FIG. 7 shows the sampling point correcting circuit 28 and its peripheral circuit The sampling point correcting circuit 28 supplies an output pulse from a sampling clock generator 61 as a sampling clock to the A/D converter 24 shown in FIG. 3 via a voltage-controlled pulse delay circuit 62. A delay time is determined by the computer 12 by the above-mentioned procedure and supplied as a control voltage from the computer 12 to the pulse delay circuit 62 via an interface 63 and a D/A converter 64. A delay time td is determined to satisfy, e.g., 0.ltoreq.td<.DELTA.ts assuming that a sampling interval is .DELTA.ts.

20 FIG. 5 shows the reference wave phase correcting circuit 26 and its peripheral circuit. In the reference signal phase correcting circuit 26, a voltage-controlled phase shifter 42 shifts the phase of a phase detecting reference signal output from a reference signal generator 41 by the correction amount .DELTA..phi.c determined by the above-mentioned method. A control voltage corresponding to .DELTA..phi.c is obtained by the computer 12 and supplied to the voltage-controlled phase shifter 42 via an interface 43 and a D/A converter 44. The range of .DELTA..phi.c is, e.g., .vertline..DELTA..phi.c.vertline..ltoreq..pi..

21 In the above embodiment, the description has been made assuming that a 90.degree. relative phase of quadrature detection and amplitude balance between channels are adjusted beforehand. However, by using hardware and a correction procedure substantially the same as those in the above absolute phase correction, an offset from an ideal state of one or both of the 90.degree. relative phase of quadrature detection and the amplitude balance between channels can be automatically corrected.

22 In addition to the above conventional method such as the Fourier method or the spin warp method as described above, the present invention can be applied to ultra-high-speed imaging such as an echo planar method based on a sequence shown in FIG. 10. In this case, base line correction, sampling point correction, and reference signal phase correction are performed for all signals (e.g., an FID, a first echo, and a second echo) shown in FIG. 10, if necessary. This correction can be performed by hardware and a procedure similar to those in the above embodiment except that a correction value must

be sequentially changed as shown in FIG. 10 by the system controller 10 shown in FIG. 2 because correction values are generally different for the respective signals (assuming that a phase error of a kth (k=1, 2, 3, . . .) echo is .DELTA..phi.k, a shift amount of a sampling point is .DELTA.tsk, and a base line amount is Vbk(t), .DELTA..phi.1.noteq..DELTA..phi.2.noteq..DELTA..phi.3 . . . , .DELTA.ts1.noteq..DELTA.ts2.noteq..DELTA.ts3 . . . , Vb1(t).noteq.Vb2 (t).noteq.Vb3(t) . . .). When half encoding is applied to the echo planar method shown in FIG. 10, data acquisition and conjugate data generation in the Fourier space are as shown in FIG. 9(b).

23 Such a correction method is effective especially in a real-time magnetic resonance imaging system capable of repeating the sequence shown in FIG. 10 at high speed to perform real-time imaging of a dynamic organ such as a heart. That is, when the above correction is applied to the real-time magnetic resonance imaging system, the correction as shown in FIG. 2 conventionally performed by software can be omitted. Therefore, image reconstruction can be performed by only the two-dimensional Fourier transform in real time. Note that in the real-time magnetic resonance imaging, a .theta..degree. pulse (0<.theta..degree.<90.degree.) corresponding to a small flip angle must be used in place of a 90.degree. pulse as an RF pulse. In addition to the above embodiment, the correcting circuits according to the present invention can be applied to any system, e.g., multi-slice imaging, three-dimensional magnetic resonance imaging, and chemical shift imaging. Furthermore, the present invention can be variously modified and carried out without departing from the spirit and scope of the invention.

24 It is a matter of course that even when only some of the base line correcting circuit, the sampling point correcting circuit, and the reference signal phase correcting circuit are used, the corresponding software correction processing can be omitted to reduce a load on software of a system.

25 Industrial Applicability

26 According to the present invention, since a base line correcting circuit for correcting a base line of a magnetic resonance signal, a sampling point correcting circuit for correcting a sampling point of the magnetic resonance signal, and a reference signal phase correcting circuit for correcting a phase of a phase-sensitive detection reference signal of the magnetic resonance signal are provided, image reconstruction can be performed by only a Fourier transform operation using a computer. Therefore, there is provided a magnetic resonance imaging system having a very high diagnostic efficiency and a more practical real-time magnetic resonance imaging system.

27 In addition, by providing at least some of the base line correcting circuit for correcting a base line of a magnetic resonance signal, the sampling point correcting circuit for correcting a sampling point of the magnetic resonance signal, and the reference signal phase correcting circuit for correcting a phase of a phase-sensitive detecting reference signal of the magnetic resonance signal according to the present invention, pre-processing performed by a computer for image reconstruction can be reduced. Therefore, a magnetic resonance imaging system having a high diagnostic efficiency can be provided.

CLAIMS:

I claim:

1. A magnetic resonance imaging system which includes magnetic resonance signal detecting means (1, 2, 3, 4, 7, 8, 10) for applying pulses of an RF

magnetic field and slicing, phase-encoding, and reading gradient magnetic fields to an object to be examined placed in a uniform static magnetic field in accordance with a predetermined sequence, thereby detecting a magnetic resonance signal from said object to be examined, receiving means (9) for phase-sensitive detecting and amplifying the magnetic resonance signal, data acquiring means (11) for sampling and digitizing the magnetic resonance signal obtained by said receiving means (9), and image reconstructing means (12) for performing image reconstruction on the basis of sampling data of the magnetic resonance signal obtained by said data acquiring means (11), comprising:

a reference signal phase correcting circuit (26) for correcting a phase of a phase-sensitive detecting reference signal in said receiving means (9);

a base line correcting circuit (27) for correcting a base line of the magnetic resonance signal detected by said magnetic resonance signal detecting means (1, 2, 3, 4, 7, 8, 10); and

a sampling point correcting circuit (28) for correcting a sampling point of the magnetic resonance signal in said data acquiring means (11).

2. A system according to claim 1, wherein said reference signal phase correcting circuit (26), said base line correcting circuit (27), and said sampling point correcting circuit (28) are controlled on the basis of a phase correction value, a base line value, and a sampling point shift amount, respectively, obtained by a computer (12) prior to image reconstruction.

3. A system according to claim 1, wherein said base line correcting circuit (27) includes means for controlling an offset value of an amplifying system of said receiving means (9).

4. A system according to claim 1, wherein said reference signal phase correcting circuit (26), said base line correcting circuit (27), and said sampling point correcting circuit (28) are controlled on the basis of a phase correction value, a base line value, and a sampling point shift amount, respectively, obtained by a computer (12) prior to image reconstruction in accordance with a sequence not performing a phase encoding beforehand.

5. A system according to claim 1, wherein said magnetic resonance signal detecting means (1, 2, 3, 4, 7, 8, 10) includes means for obtaining only magnetic resonance data corresponding to Fourier data in a half region of a Fourier space, and said image reconstructing means (12) includes means for obtaining data of the other half space on the basis of the magnetic resonance data by utilizing the fact that data point-symmetrical about the origin of the Fourier space are complex conjugate with each other, thereby performing image reconstruction.

6. A system according to claim 1, wherein said magnetic resonance signal detecting means (1, 2, 3, 4, 7, 8, 10) includes means for detecting a multiple echo signal train as a magnetic resonance signal, and said reference signal phase correcting circuit (26) includes means for independently correcting the phase-sensitive detecting reference signal for each echo signal of the multiple echo signal train.

7. A system according to claim 1, wherein said magnetic resonance signal detecting means (1, 2, 3, 4, 7, 8, 10) includes means for detecting a magnetic resonance signal generated upon magnetic resonance excitation of a flip angle of less than 90.degree..

8. A magnetic resonance imaging system which includes magnetic resonance signal detecting mans for applying pulses of an RF magnetic field and slicing, phase-encoding, and reading gradient magnetic fields to an object to be examined placed in a uniform static magnetic field in accordance with a predetermined sequence, thereby detecting a magnetic resonance signal from said object to be examined, receiving means for phase-sensitive detecting and amplifying the magnetic resonance signal, data acquiring means for sampling and digitizing the magnetic resonance signal obtained by said receiving means, and image reconstructing means for performing image reconstruction on the basis of sampling data of the magnetic resonance signal obtained by said data acquiring means, comprising:

a base line correcting circuit for correcting a base line of the magnetic resonance signal detected by said magnetic resonance signal detecting means; and

a sampling point correcting circuit for correcting a sampling point of the magnetic resonance signal in said data acquiring means.

9. A system according to claim 8, wherein said base line correcting circuit and said sampling point correcting circuit are controlled on the basis of a base line value and a sampling point shift amount, respectively, obtained by a computer prior to image reconstruction.

10. A system according to claim 8, wherein said base line correcting circuit includes means for controlling an offset value of an amplifying system of said receiving means.

11. A system according to claim 8, wherein said base line correcting circuit and said sampling point correcting circuit are controlled on the basis of a base line value and a sampling point shift amount, respectively, obtained by a computer prior to image reconstruction in accordance with a sequence not performing a phase encoding beforehand.

12. A system according to claim 8, wherein said magnetic resonance signal detecting means includes means for obtaining only magnetic resonance data corresponding to Fourier data in a half region of a Fourier space, and said image reconstructing means includes means for obtaining data of the other half space on the basis of the magnetic resonance data by utilizing the fact that data point-symmetrical about the origin of the Fourier space are complex conjugate with each other, thereby performing image reconstruction.

13. A system according to claim 8, wherein said magnetic resonance signal detecting means includes means for detecting a multiple echo signal train as a magnetic resonance signal.

14. A system according to claim 8, wherein said magnetic resonance signal detecting means includes means for detecting a magnetic resonance signal generated upon magnetic resonance excitation of a flip angle of less than 90.degree..

Previous Doc Next Doc Go to Doc#

L18: Entry 19 of 19 File: USPT Aug 15, 1989

US-PAT-NO: 4857846
DOCUMENT-IDENTIFIER: US 4857846 A

TITLE: Rapid MRI using multiple receivers producing multiply phase-encoded data derived from a single NMR response

DATE-ISSUED: August 15, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Carlson; Joseph W.	San Francisco	CA		

ASSIGNEE-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY	TYPE CODE
The Regents of the University of California	Oakland	CA			02

APPL-NO: 07/ 176251 [PALM]
DATE FILED: March 31, 1988

INT-CL: [04] G01R 33/20

US-CL-ISSUED: 324/309; 324/314
US-CL-CURRENT: 324/309; 324/314

FIELD-OF-SEARCH: 364/324.05, 324/300, 324/307, 324/309, 324/311, 324/312, 324/314, 324/313

PRIOR-ART-DISCLOSED:

U.S. PATENT DOCUMENTS

Search Selected Search ALL Clear

PAT-NO	ISSUE-DATE	PATENTEE-NAME	US-CL
4418316	November 1983	Young et al.	324/309
4598368	July 1986	Umemura	364/324.05
4611172	September 1986	Takase	324/314
4616181	October 1986	Kemner	324/309
4652827	March 1987	Eguchi	324/312
4668915	May 1987	Daubin et al.	324/309

☐	4682112	June 1987	Beer	
☐	4719581	January 1988	Sakamoto	324/307
☐	4727327	February 1988	Toyoshima et al.	324/309
☐	4746865	May 1988	Kasugai	324/309

OTHER PUBLICATIONS

Hutchinson et al, "Fast MRI Data Acquisition Using Multiple Detectors," Mag. Res. Med. 6, 87-91 (1988).
Kumar A., Welti D. and Ernst R. R., NMR Fourier Zeugmatography, Journal of Magnetic Resonance, 1975; 18:69-83.
Sutherland R. J. and Hutchinson J. M. S., Three-Dimensional NMR Imaging Using Selective Excitation, Journal of Physics E 1978; 11:79-83.
Lauterbur P. C., Image Formation by Induced Local Interactions: Examples Employing Nuclear Magnetic Resonance, Nature 1973; 16:242-243.
den Boef F. H., van Uijen C. M. T. and Holzscherer C. D., Multiple-Slice NMR Imaging by Three-Dimensional Fourier Zeugmatography, Physics in Medicine and Biology 1984; 29:857-867.
Feinberg D. A., Hale J. D., Watts J. D., Kaufman L. and Mark A., Halving ME Imaging Time by Conjunction: Demonstration at 3.5KG, Radiology 1986; 161:527-531.
Ernst R. R., Sensitivity Enhancement in Magnetic Resonance in: Waugh J. S., ed., Advances in Magnetic Resonance, vol. 2, N.Y.: Academic Press, 1966: 1-135.
Carlson J., Crooks L. E., Ortendahl D. A., Kramer D. M. and Kaufman L., Technical Note: Comparing S/N and Section Thickness in 2-D and 3-DFT MRI, Radiology 1988; 166:266-270.

ART-UNIT: 265

PRIMARY-EXAMINER: Tokar; Michael J.

ATTY-AGENT-FIRM: Nixon & Vanderhye

ABSTRACT:

Method and apparatus for more rapidly capturing MRI data by receiving and recording NMR RF responses in plural substantially independent RF signal receiving and processing channels during the occurrence of an NMR RF response. The resulting plural data sets respectively provided by the plural RF channels are then used to produce multiply phase-encoded MRI data from the single NMR RF response. Practical examples are disclosed for reducing required MRI data capturing time by factors of at least about one-half.

23 Claims, 15 Drawing figures
Exemplary Claim Number: 1
Number of Drawing Sheets: 6

BRIEF SUMMARY:

1 This invention is generally related to magnetic resonance imaging (MRI) using nuclear magnetic resonance (NMR) phenomena. It is particularly directed to method and apparatus for more efficiently capturing and providing MRI data suitable for use in multi-dimensional Fourier transformation MRI imaging processes.

2 MRI is by now a widely accepted and commercially viable technique for obtaining digitized video images representative of internal body structures. There are many commercially available approaches and there have been numerous publications describing these and other approaches to MRI. Many of these use multi-dimensional Fourier transformation techniques which are, by now, well-known to those skilled in this art.

3 For example, in one commercially available MRI system, a slice selective G.sub.z magnetic gradient pulse is utilized in conjunction with RF nutation pulses (including a 180.degree. nutation pulse) to produce true spin echo NMR RF pulses from only a relatively narrow "planar" or "slice" volume perpendicular to the Z axis. During readout and recordation of the NMR spin echo RF response, a G.sub.x magnetic gradient pulse is employed to provide spatially dependent frequency/phase-encoding in the X axis dimension. Accordingly, by a first one-dimensional Fourier Transformation process, one can obtain Fourier coefficients representing the NMR spin echo response at different X locations emanating from a correspondingly located "column" volumes parallel to the Y axis. By rapidly repeating this same process using different slice selective G.sub.z gradient pulses during a single T1 NMR interval, it is known that one can significantly enhance the efficiency of obtaining data for a number of planar volumes (sometimes called "multi-slice" MRI).

4 However, one cannot produce the requisite two-dimensional visual image from only a single dimension of Fourier transformation (per slice) as just described. To obtain the second dimension of Fourier coefficients for resolution in the Y axis dimension, a different G.sub.y phase-encoding pulse (e.g., different in magnitude and/or time duration) is utilized during the NMR excitation process such that the NMR spin echo responses during different data gathering cycles will produce Fourier coefficients phase-encoded with respect to spatial location in the Y axis dimension.

5 Accordingly, if one wants to obtain, for example, a resolution of 256 pixels along the Y axis dimension, then one must go through 256 data gathering cycles (per slice) with correspondingly different Y axis phase-encoding in each cycle to assemble the requisite data required for the second dimension of Fourier transformation.

6 Because the normally encountered T1 NMR parameter is on the order of a second or so in many human tissues, and because one typically does not (in this exemplary system) repeat a data taking cycle within the same volume until the previous NMR nuclei have substantially returned to their quiescent conditions, it will be appreciated that the need to repeat a data taking cycle literally hundreds of times translates into an overall MRI data capturing process that requires several minutes to complete. And some of the more interesting future MRI applications may even require thousands of data taking cycles and several tens of minutes using current technology. At the same time, as is well known in the art, MRI systems, facilities and operating personnel represent a significant expense which can only be economic if the time required for capturing NMR data per patient is minimized.

7 In addition to the rather basic economic improvement that can be obtained by reducing MRI data capture time, it may also make practical some rather sophisticated new MRI possibilities. For example, three dimensional MRI may be useful for permitting one to obtain oblique reconstruction images on any desired (oblique) plane. For this application, one should use isotropic

resolution which implies the creation of an MRI data set having 256 voxels (volume picture elements) in each of three mutually orthogonally dimensions. Even if partial flip imaging techniques are used to minimize intervals between NMR excitation sequences, current MRI techniques might still require imaging times on the order of a half hour--which is probably beyond a reasonable imaging time for most patients. If this time requirement could be reduced by at least a factor of one-half, it might become a much more attractive possibility.

8 Similarly, echo planar imaging techniques (which permit all data required for a single image to be obtained after a single excitation) would become more practical if higher resolution data could be obtained during the ensuing train of NMR RF responses (which is necessarily limited by the T2 decay parameter).

9 The enhanced diagnostic possibilities that may someday be provided by spectroscopic MRI imaging might also become more practical if some technique is developed for shortening the time required to collect requisite spectroscopic imaging data. MRI angiography is another technique which presently requires unusually long MRI imaging times and which may become considerably more practical if a reduced imaging time technique could be employed.

10 Furthermore, it is perhaps self-evident that motion artifact can be reduced if one can somehow shorten the time interval over which requisite MRI data is collected.

11 In short, standard techniques of two-dimensional Fourier transform magnetic resonance imaging are already highly efficient in two out of three dimensions. Through the use of selective excitation and data acquisition with a readout gradient, the Fourier transform of the acquired data effectively localizes signal in two dimensions. However, there remains a time consuming aspect of MRI insofar as it is still necessary to obtain multiple phase-encoded data acquisitions to localize the signal in the third (Y axis) dimension in many standard MRI processes.

12 The need for reducing MRI image time requirements has been recognized by many others. There are other techniques for more rapidly obtaining requisite MRI data. For example, other techniques for rapid MRI imaging have attempted to reduce the time spent on acquiring phase-encoded acquisitions. For example, data conjugation techniques (exploiting the symmetries of the Fourier transform) are already employed in some commercial processes to reduce the number of required acquisitions by a factor of two. Partial flip angle nutation (instead of a full 90.degree. initial RF nutation angle) may also be employed to eliminate signal loss normally encountered with rapid acquisitions (e.g., as in short repetition time techniques). In principle, the most efficient means of signal acquisition for MRI may involve the use of only partial initial RF nutation angles or "partial flip" combined with a very short TR interval between repetitions of the NMR excitation-response processes used in the data gathering phase of MRI. For example, this latter partial flip technique has aided in the development of three dimensional Fourier transform techniques for imaging a large number of sections within a very short TR.

13 Some publications generally relevant to such MRI techniques as have just been discussed may be noted as follows:

14 Kumar A., Welti D. and Ernst R. R. NMR Fourier Zeugmatography. Journal of Mangetic Resonance 1975; 18:69-83.

15 Sutherland R. J. and Hutchison J. M. S. Three-Dimensional NMR Imaging Using Selective Excitation. Journal of Physics E 1978; 11:79-83.

16 Lauterbur P. C. Image Formation By Induced Local Interactions: Examples Employing Nuclear Magnetic Resonance Nature 1973; 16:242-243.

17 den Boef J. H., van Vijen C. M. T. and Holzscherer C. D. Multiple-Slice NMR Imaging by Three-Dimensional Fourier Zeugmatography. Physics in Medicine and Biology 1984; 29:857-867.

18 Feinberg D. A., Hale J. D., Watts J. C. Kaufman L. and

19 Mark A. Halving MR Imaging Time By Conjugation: Demonstration at 3.5KG. Radiology 1986; 161:527-531.

20 Ernst R. R. Sensitivity Enhancement in Mangetic Resonance In: Waugh J. S., ed. Advances in Mangetic Resonance Vol. 2, N.Y.: Academic Press, 1966:1-135.

21 Carlson J., Crooks L. E., Ortendahl D. A., Kramer D. M. and Kaufman L. Technical Note: Comparing S/N and Section Thickness in 2-D and 3-DFT MRI. Radiology 1988;166:266-270

22 I have discovered a new MRI reconstruction algorithm which permits a reduction in the number of required phase-encoding data acquisitions in an MRI imaging sequence without reducing either resolution or field of view. One principle of this technique involves the use of two (or more) non-interacting receiver coils (and corresponding independent RF receiving and signal processing channels) so as to each simultaneously detect an NMR response signal from the same tissue. An imaging sequence may use phase-encoded spin echoes or gradient echoes in the usual way; however, the reconstruction algorithm effectively calculates multiple Fourier projections of the tissue from but a single NMR response (e.g., a spin echo).

23 A full implementation of this discovery using but two receiver coils (and associated RF signal processing channels) can reduce the number of needed phase-encoded data acquisitions by 50% as compared to the fewest number otherwise (e.g,. previously) required.

24 In essence, my new technique is a variation of standard two-dimensional or three-dimensional MRI but, instead of using a single RF receiver coil and its associated RF signal processing channel, I use at least two (or more) different non-interacting receiver coils and associated RF signal processing channels to simultaneously detect an NMR response signal emanating from a given tissue. The coils are constructed and/or oriented so that their respective responses to the same NMR signal are different--i.e., the coil responses respectively depend upon the relative spatial location of the nuclei emitting the NMR RF response. This added spatial dependency provides additional information from the same single NMR RF response which can be effectively used to reduce the number of NMR data acquisition cycles required for a given resolution of MRI image reconstruction. In short, the MRI reconstruction algorithm is permitted to calculate multiple Fourier projections from but a single NMR RF response (e.g., a single spin echo) resulting in a significant time savings in required MRI data acquisition time.

25 Some aspects of my invention have already been published ("An Algorithm for NMR Imaging Reconstruction Based on Multiple RF Receiver Coils," Journal of

Magnetic Resonance J. Mag. Res., Vol 74, pp 376-380, 1987). And others, subsequent to my invention, have now proposed the general concept of using multiple detectors for MRI data acquisition so as to save data acquisition time--but without any apparent practical implementations in mind (see Hutchinson et al, "Fast MRI Data Acquisition Using Multiple Detectors," Magnetic Resonance in Medicine, Vol. 6, 1988, pp 87-91).

26 This invention is especially useful in MRI data capturing sequences where a dimension transverse to the static magnetic field is phase-encoded over a plurality of data gathering sequences to obtain requisite MRI data for a single image. It may be used with both two and three-dimensional Fourier Transform Magnetic Resonance Imaging processes. Preferably, in the exemplary embodiments, the static magnetic field is horizontal (as in a super conducting solenoidal magnet) so that the exemplary RF receiving coils permit the most convenient patient access.

27 In this invention, a system of multiple RF receiving coils provides some localization of NMR RF signal responses in at least the phase-encoded dimensions. By using spatial dependence of the phase (primarily) and the amplitude (secondarily) of the NMR RF response signal induced in a set of non-interacting RF receive coils, it is possible to calculate multiple phase-encoded signals from but a single NMR RF response (e.g., a single NMR spin echo response). This technique is compatible with existing two and three dimensional imaging techniques and may also find use in conjunction with other rapid MRI imaging techniques. The exemplary embodiments provide for a reduction in the required minimum number of NMR RF responses by a factor of at least about two.

28 It is worth noting some specific imaging protocols to which this rapid technique may be applied. These examples are not intended to be comprehensive, but rather, protocols chosen to illustrate sequences which are presently limited by the number of necessary phase-encoded acquisitions:

29 (1) 3D partial flip imaging can achieve a realistic minimum TR of approximately 50 msec. Below this value data acquisition time, i.e., echo length, must be decreased, resulting in an increased bandwidth with a consequent increase in noise. One promising utility of 3D MRI is oblique reconstruction to form images on any plane. For this one should use isotropic resolution which would imply a high resolution data set of 256.sup.3 voxels. Using a 50 msec TR, this gives a minimum imaging time of 55 minutes. Data conjugation may reduce this to 27 minutes, but this is still beyond reasonable imaging time for most patients. A reduction to 14 minutes through the use of multiple receivers makes this an attractive possibility.

30 (2) Echo planar imaging collects all data for a single image after one excitation. Data acquisition time is therefore limited by T2 of the tissue. Presently resolution is limited by the number of projections available within this time. One laboratory's limitation on resolution is 128 phase-encoded projections. Multiple receiver coil reconstruction will allow for higher resolution (e.g., 256 projection) echo planar imaging.

31 (3) Short TR partial flip imaging has been seen to be inappropriate for some imaging protocols. Tissue contrast generally decreases in short TR partial flip images, which can be a disadvantage in some diagnoses. Metallic implants and magnetic field inhomogeneities give rise to artifacts that further degrade the images. (Since it relies on gradient echoes, 2D partial flip MRI is much more susceptible to these artifacts.) Multiple receiver coil reconstruction may allow for new flexibility in protocols which manipulate image contrast in

rapid scan.

32 Further expansion of the ability of MRI to serve clinical needs relies on reduction of imaging time. This new approach towards rapid MRI can be used with existing methods and can further reduce imaging time. Other possible applications involve imaging times longer than those of standard MRI. For example, spectroscopic imaging is hindered in patient acceptance by the very long imaging times required. A reduction by 50% in imaging time can certainly improve its clinical usefulness. MRI angiography generally requires two to four times the usual imaging time. In this case, a reduction in time will aid not only in acceptance, but also in improving the image registration by reducing patient motion during the scan time. Motion artifacts produced by respiratory motion decrease with signal averaging. Thus, multiple coil image reconstruction may be used as a way to reduce artifacts in a fixed imaging time. Multiple coil reconstruction will add to the flexibility of implementing many MRI procedures.

33 These examples are described to illustrate how a reduction in phase-encoded data acquisition time can provide new possibilities in imaging protocols. A distinct use of this reconstruction is as a technique that increases the number of data set acquisitions effected in the same amount of time. While this will probably not result in an improvement in the signal to noise ratio, it may be useful in reducing motion artifacts. A discussion of signal to noise in the exemplary embodiment will be presented later.

34 Further improvements can be had by adding additional receiver coils. In general one may consider a set of N receivers. Using the exemplary reconstruction algorithm, one may then calculate N phase encoded projections per echo. Introducing more receiver coils adds to the technical difficulty of implementation due to the necessity of maintaining electrical isolation between the receiving coils.

35 In one exemplary embodiment, a pair of essentially co-located "birdcage" coils is utilized. One of the coils is tuned to a fundamental frequency corresponding to the desired NMR frequency band (e.g., about 15 MHz) while the second coil is tuned so that its second harmonic is at the same frequency (e.g., about 15 MHz). As will be shown in more detail below, this results in the requisite current/voltage distribution on the axially extending coil wires to take on the form of sin .theta. and sin 2.theta., respectively, where .theta. is a relative wire location angle in the x,y plane. Standard MRI RF coil construction techniques are used to minimize mutual inductance or capacitive coupling between the two coils so as to keep them substantially independent of one another. Saddle coil constructions may also be utilized to obtain the requisite respective sin .theta. . . . sin N.theta. current/voltage distributions in the N receiving coils.

DRAWING DESCRIPTION:

These as well as other objects and advantages of this invention will be more completely understood and appreciated by carefully reading the following detailed description of the presently preferred exemplary embodiments in conjunction with the attached drawings, of which:

FIG. 1 is an abbreviated schematic block diagram of an MRI system employing this invention;

FIG. 2 is a schematic arrangement showing an arrangement of multiple RF receiver coils useful in explaining the underlying theory of this invention;

FIG. 3 is a schematic depiction of a cross-section of the multiple coil construction shown in FIG. 2;

FIG. 4 is a schematic view of one possible practical embodiment of two co-located saddle coils for implementing the simplest multiple coil case where N=1;

FIGS. 5A 5B, 5C, and 5D are further schematic depictions of various birdcage and saddle coil constructions that might be utilized for achieving a practical RF receive coil for use with this invention;

FIG. 6 is a graph showing spin density of a simple one-dimensional sample;

FIG. 7 is a graph depicting standard Fourier reconstruction of the samples shown in FIG. 6;

FIG. 8 depicts a two-coil reconstruction of a point sample using three projections per echo;

FIG. 9 depicts a two-coil reconstruction of a point sample using two projections per echo;

FIG. 10 depicts a two-coil reconstruction of a point sample using an averaged value of two projections per echo;

FIG. 11 depicts a standard Fourier reconstruction of the same sample with Gaussian white noise added to the sample signal; and

FIG. 12 depicts a two-coil reconstruction of the same sample with sample generated noise.

DETAILED DESCRIPTION:

1 In FIG. 1 an MRI system 100 is schematically depicted as including the usual static magnet, gradient coils, shim coils, transmit RF coils, 102 under control of processor 104 (which typically communicates with an operator via a conventional keyboard/control display module 106). Although, it is conceivable that a single processor might both control the system and also carry out the actual MRI imaging processes, it is perhaps more conventional to employ a system of multiple processors for carrying out specialized functions within the MRI system 100 as will be appreciated. Accordingly, as depicted in FIG. 1, an MRI image processor 108 receives digitized data representing RF NMR responses from an object under examination (e.g., a human body 110) and, typically via multiple Fourier transformation processes well-known in the art, calculates a digitized visual image (e.g., a two-dimensional array of picture elements or pixels, each of which may have different gradations of gray values or color values, or the like) which may then be conventionally displayed at 112.

2 In accordance with this invention, a plurality of receive coils 1 . . . N are independently coupled to a common imaging volume (e.g., a desired portion of body 110). The RF signals emanating from these coils are respectively processed in independent RF channels 1 through N. As depicted in FIG. 1, each such RF channel may comprise a considerable amount of conventional analog RF signal processing circuits as well as an eventual analog to digital conversion before being input to the MRI processor 108 (which may typically include means for digitally storing the acquired data during a data acquisition sequence until the image processor 108 uses such acquired data to produce an image at 112).

3 Although the RF channel circuitry may typically include a rather complex (and expensive) amount of circuitry, it is not believed necessary to describe it in any detail since conventional RF signal processing circuitry per se may be employed with this invention. However, the extra expense of using extra receive coils and associated RF signal processing channels does have to be balanced against the improvement in data capturing time when considering the overall economics of this approach.

4 Accordingly, to practice this invention, modifications need be made in essentially only three areas of a conventional MRI processing system:

5 1. additional receive coils need to be employed and their construction is preferably such that they are substantially independent (i.e., effectively without substantial mutual inductive coupling or capacitive coupling);

6 2. an additional RF signal processing channel needs to be added for each of the additonal receive coils; and

7 3. the MRI reconstruction algorithm programmed into and implemented by the image processor 108 needs to be slightly modified so as to use the additional incoming data to calculate multiply phase-encoded MRI data in an appropriate way.

8 Since the RF signal processing channels are, per se, simply replications of existing conventional RF channels, it is not believed that any further detailed description of such additional channels need be given in this application. Rather, the following disclosure will concentrate on practical exemplary embodiments for the receive coil and for the new calculations to be made in the image processor 108.

9 Present reconstruction techniques in two dimensional NMR imaging are highly efficient in two of the three dimensions. One limitation is the large number of phase-encoded echoes which must be acquired in order to reconstruct a full image. In an attempt to facilitate the development of very fast NMR imaging techniques, it is useful to consider reconstruction techniques which do not rely on multiple acquisitions of phase-encoded spin echo signals.

10 One approach for reducing the number of phase-encoded spin echoes required for a reconstruction of an image is described below. It uses multiple coils for the acquisition of signals, then uses the resulting additional information to speed the reconstruction process. The reconstruction algorithm is, initially, in the following discussion, based on an idealized NMR detector. Thus, theoretically, one can conceive a situation in which it may be possible to reconstruct an entire image from but a single NMR spin echo! However, as one might expect, realities of signal-to-noise make this a highly unrealistic situation. Nonetheless, the description is presented since it is the easiest way to initially demonstrate the algorithm. A more practical implementation of this algorithm is described later in which a set of two coil "building blocks" is used to reduce the number of required spin echoes by a factor of two.

11 As an introduction to the technique, consider the idealized NMR detector in FIG. 2. Long, straight wires 200 (parallel to the static magnetic field H.sub.o) run along the surface of a (non conductive) cylinder 202. The ends of these loops are closed at infinity. Only a few wires are shown. In the idealized coil, wires would densely surround the cylinder. Assume that the voltages on each of these loops induced by a precessing magnetic NMR dipole can be monitored separately. Denote the voltage induced on the loop whose wire

is at an angle .theta. from the vertical by V(.theta.).

12 Standard two dimensional reconstruction allows a straightforward procedure for localizing magnetization in two dimensions by means of slice selection and read-out gradients (G.sub.z and G.sub.x, respectively). The time consuming processes are the number of required repetitions of the basic cycle to acquire the requisite phase-encoded data for resolving nuclei along the y-axis dimension. The problem at hand is to devise an algorithm to reconstruct a column 300 of magnetization perpendicular to the slice selection and read-out directions (e.g., along the y-axis). A schematic depiction of the situation is shown in FIG. 3. A column of magnetization 300 is centered at location x and z (z can be taken to be zero). Imagine that one is reconstructing an image of a column 300 as shown.

13 The voltage induced in the loop whose inner wire is located at angle .theta. is proportional to the magnetic field produced by that wire if it were a transmitter and driven with unit current. The magnetic field B.sub.x +iB.sub.y at location (x,y) due to the wire at .theta. is: ##EQU1## The voltage induced on the wire at .theta. can then be written as a superposition of the voltages induced by all magnetization: ##EQU2## with .alpha.=e.sup.i.theta.. Overall constants have been ignored.

14 This can easily be inverted with the contour integral ##EQU3## The integration contour is the unit circle. Since the integrand has a simple pole at .alpha.=i (x-iy)/R with residue -ie.sup.-n.pi.(x-iy)/R /R, the integral is easily performed. The answer is: ##EQU4## Since x is a known quantity, one can see that the result of the integration gives a value for the Fourier Transform of the magnetization density.

15 A crucial observation is that a complete determination of V(.theta.) together with the evaluation of this integral yields all values of the Fourier Transform of the magnetization density. However, if the magnetization has experienced a phase-encoding gradient previously in the sequence with a strength G.sub.y and duration .tau., then the magnetization density along a column is

m(x,y)e.sup.i.gamma.GyY.tau.. [Equation 5]

16 In this case, evaluation of the integral yields values of the Fourier Transform at different values in k space.

17 One difficulty in evaluating the integral in the reconstruction arises from the exponential term in the integrand:

e.sup.in.pi.e.spsp.u.theta.. [Equation 6]

18 When .theta.=-.tau./2, the exponential term is e.sup.+n.pi.. For n larger than 3 or 4, the size of this peak makes an accurate evaluation of the integral difficult.

19 One way to circumvent the difficulties in the numerical integration is to presume in the idealized NMR receiver that one is able to measure the Fourier Transform of V(.theta.). That is, what is measured are the coefficients .alpha..sub.k from the Fourier expansion of the voltage: ##EQU5## Substituting this expression into the reconstruction integral: ##EQU6## The integral over .alpha. vanishes unless k is negative. This result gives the answer: ##EQU7## In other words, by measuring the negative Fourier coefficients of the voltage induced in the wires and performing a sum one may

calculate values of the Fourier Transform of the magnetization density.

20 Now it is time to discuss a more realistic implementation. In an actual embodiment, it is likely that multiple receiver coils would be used to gather sufficient data to reconstruct two (or possibly three or more) values of the Fourier Transform of the magnetization density. Standard phase-encoding pulses would still be needed in order to gather all projections. Consider the relatively simple coil of four wires. The first negative Fourier coefficient of the voltages is given by: ##EQU8##

21 The expression has been rewritten to make explicit the following fact: the term V(O)-V(.pi.) is nothing but the voltage induced in a loop that has one wire along the cylinder at .theta.=0 and a return path along the bottom of the cylinder at .theta.=.pi.. So also, the other two terms in the voltage Fourier coefficient are the same as the voltage induced in the horizontal loop.

22 This observation yields an actual practical implementation. The basic "building block" coil is a two loop arrangement of FIG. 4. From this, the voltage Fourier coefficient .alpha..sub.-1 is given by the voltage induced in the vertical saddle coil loop 1 combined with the voltage in the horizontal saddle coil loop 2 shifted by 90 degrees. This can be viewed as a simple quadrature receiver. In fact, it can be implemented as a four legged birdcage coil receiving in quadrature.

23 The next question is whether one may measure additional Fourier coefficients from this four wire situation. The answer is no. Given only four wires, the next negative Fourier coefficient, .alpha..sub.-2, is indistinguishable from the Fourier coefficient .alpha..sub.2. In order to measure higher Fourier coefficients one must have additional loops.

24 A straightforward way to do this is to use an eight wire configuration. One may construct the voltage Fourier coefficients .alpha..sub.-1 and .alpha..sub.-2 from the signals induced in each wire on the cylinder. That is, ##EQU9## This can be recognized as the signal in a building block coil combined with the 45 degree phase shift of the signal on the second building block coil. Again, one implementation can use a birdcage coil receiving in quadrature. This time the receiver is an eight legged cage.

25 The voltage Fourier coefficient .alpha..sub.-2 is written in a similar form: ##EQU10## This is not a simple superposition of signals from building block coils. Instead, it is the signal of two opposed saddle coils receiving in quadrature. One way to realize this coil is to construct a quadrature eight legged birdcage coil tuned to its second harmonic resonant frequency. Of course, the coil must be distinct from the first in order for its second resonant frequency to coincide with the first resonant frequency of the first.

26 Higher order situations are similarly constructed.

27 Two major questions remain concerning this technique: (1) what is the signal to noise cost in the image and (2) how can one construct coils which behave as independent receivers. Simulations of data reconstruction have been performed using eight wire coils to reduce the number of phase-encoding acquisitions by a factor of two. The response to noise generated by the sample--and hence correlated in the coils--presently appears to be identical in this reconstruction and in standard two dimensional imaging. A serious concern is that reconstruction is impossible if noise generated by the coils is sufficiently high.

28 Coil coupling is another concern. Only if the coils can be substantially decoupled can the reconstruction proceed in the manner described previously. The implementation using two birdcage coils offers a possibility of solving the coupling problem since two birdcage coils operating in different modes (but the same frequency) have no intrinsic mutual inductance between coils. And capacitive coupling may be minimized (e.g., using techniques employed for realizing practical quadrature detection coils, per se).

29 As will be appreciated by those in the art, practical coil constructions for RF receiving in MRI typically include tuning and impedance matching capacitances in conjunction with transmission lines, etc. For purposes of simplifying the discussion in this case, such conventional aspects of RF receiving coils and their associated RF transmission lines, etc., are not depicted in the FIGURES or otherwise described.

30 Another way to understand the underlying theory, is to imagine a simplified imaging experiment which attempts to reconstruct the transverse magnetization density in a sample tube that is parallel to the y axis. The x and z location of the column are known. What is needed, is a reconstruction of the sample along the y direction. This simplified experiment can be viewed as merely part of a standard two-dimensional Fourier Transformation MRI imaging sequence. A slice/selective RF excitation provides a section of sample with a known value of z and the read-out gradient G.sub.x provides frequency encoding in the x-direction. Thus, after Fourier Transformation of the acquired data, the Fourier content of the data in a standard two-dimensional Fourier transform acquisition provides the sum of the transverse magnetization in the hypothetical column or sample now being considered.

31 Standard two-dimensional FT imaging uses repeated acquisitions with phase-encoded echoes to generate the Fourier transform of the transverse magnetization along the y direction. If the field of view along the y direction is taken to be L, the Fourier transform is usually written as

m(n)=.intg.m(y)e.sup.2.pi.iny/NL dy [Equation 15]

32 and m=m.sub.x +im.sub.y is the complex transverse magnetization.

33 For the purposes of illustration, consider the idealized RF receiver coil (FIG. 2) for this experiment. The receiver consists of a collection of long wires parallel to the static field and laid on the surface of a cylinder. The loops are closed by return paths at infinity. In this idealized receiver coil one may measure the echo signal by monitoring the voltage induced in each wire. Denote the voltage in the wire at position .theta. by V(.theta.). As shown above, the Fourier transform of the magnetization density is expressible in terms of the Fourier transform of the induced voltages. Let V(n) be the Fourier coefficient of the voltage, then the expressions for the Fourier transform of the magnetization take the form: ##EQU11## where R is the radius of the coil; the field of view, L, is 2R.

34 Only the five lowest Fourier coefficients are shown. In principle, given sufficiently many voltage Fourier coefficients, one may construct all terms in the Fourier transform of the magnetization density. Thus, one can reconstruct the entire column of the sample within one spin echo response.

35 There are two important practical considerations that rule out the possibility of a complete reconstruction from one echo, however. One problem is coil coupling. An actual coil of separate loops surrounding a cylinder would not

behave exactly as described. One reason is that coils in receivers are tuned circuits and voltages produced by echoes induce currents. Since the idealized receiver coil has a collection of loops with large mutual inductance between loops, the signal in each loop is not that just produced by the echo. The crucial observation--and this is key to feasibility--is that even though one cannot build the idealized coil, one can build separate coils that directly measure the voltage Fourier coefficients where there is substantially no intrinsic mutual inductance between these coils.

36 One embodiment of a multiple coil set is a pair of birdcage resonators. Previously, these coils have been developed as a means of producing a homogeneous RF field. Homogeneity is due to the current distribution on the legs of a driven birdcage that varies as cos .theta.. One can use the birdcage as a quadrature transmitter and generate a complex current distribution exp(-i.theta.). In this situation, a driving voltage produces a voltage distribution across the legs of the birdcage that varies as exp(-i.theta.).

37 By using the standard reciprocity principle, we can see how to produce a coil to measure V(-1). Since the voltage distributions are the same in transmission and reception, a standard birdcage resonator produces a net output voltage proportional to the sum of voltages induced on each leg weighted by exp(-i.theta.).

38 A coil that measures V(-2) requires a birdcage coil with a current distribution that varies as cos 2.theta.. By extending the previous results one can see that this coil measures a sum of voltages induced on each let with a weighting factor of exp(-2i.theta.). One method of producing a cos 2.theta. distribution is to construct a birdcage coil and tune it to its second resonant frequency. Normally, a birdcage coil has many resonant frequencies, each of which is approximately a multiple of the lowest fundamental frequency. What one here requires is one coil whose fundamental frequency coincides with the NMR frequency (e.g., 15 MHz) and a separate coil whose second harmonic is at the same frequency.

39 The two coils may be placed concentrically around the sample with substantially no intrinsic mutual inductance. Since there is substantially no intrinsic coupling between the coils, each coil behaves independent of the other and the net signal on each output is the proper voltage Fourier coefficient. One still has to contend with capacitive coupling between the coils, as well as residual inductive coupling. However, careful adjustment of the relative alignment of the two coils and other now standard techniques for decoupling quadrature-connected receive coils should provide adequate isolation.

40 Birdcage resonators are not the only way to construct these coils. Some other configurations are shown in FIGS. 5a-5d. Note that quadrature as well as non-quadrature coils may be constructed. Non-quadrature coils will be much easier to construct, but the lower signal to noise ratio of these coils may degrade the image substantially.

41 FIG. 5a depicts a "low pass" birdcage. FIG. 5b depicts a homogeneous quadrature saddle coil. Arrows denote the direction of current flow. Since the coil has a homogeneous reception pattern, the signal is essentially the same (albeit with higher signal to noise ratio) as a non-quadrature saddle coil (FIG. 5c). Another design for a coil to measure V(-2) is shown in FIG. 5d.

42 The second practical consideration facing complete reconstruction from one echo is the signal-to-noise ratio. One expects that the signal to noise ratio

of human MRI is insufficient to allow for single echo reconstruction. Therefore, one may concentrate on a reduced implementation. For example, only two voltage Fourier coefficients will be measured in the present exemplary embodiments: V(-1) and V(-2). Given these, one can construct three Fourier coefficients of the magnetization by truncating the sums for the expression of m(n): ##EQU12##

43 An actual imaging sequence will be a hybrid technique of reconstruction using standard phase-encoding in the y direction and using the extra spatial information generated by the two voltage signals. That is, a typical data acquisition sequence may first acquire an echo with no phase encoding. From this, one can construct m(-1), m(0) and m(1). Next, a sequence will acquire an echo with a phase-encoding factor of exp(2i.pi.y/R). From this, one can construct the terms m(1), m(2) and m(3). This is repeated for the rest of the positive and all negative phase encodings:

TABLE I

Data Gathering sub-cycle	encoded magnetization values captured
a	m.sub.a (-1)
	m.sub.a (0)
	m.sub.a (1)
b	.dwnarw.
	.dwnarw.
	m.sub.b (1)
	m.sub.b (2)
	m.sub.b (3)
c	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
	m.sub.c (3)
	m.sub.c (4)
	m.sub.c (5)
.	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
.	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
.	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
	.dwnarw.
q	m.sub.q (-3)
	m.sub.q (-2)

```
                                   m.sub.q (-1)
                                       .dwnarw.
                                          .dwnarw.
                                             .dwnarw.
                                                .dwnarw.
                                                   .dwnarw.
r      m.sub.r (-5)
           m.sub.r (-4)
               m.sub.r (-3)
                   .dwnarw.
                       .dwnarw.
                           .dwnarw.
                              .dwnarw.
                                 .dwnarw.
                                    .dwnarw.
                                       .dwnarw.
           .dwnarw.
               .dwnarw.
                   .dwnarw.
                       .dwnarw.
                           .dwnarw.
                              .dwnarw.
                                 .dwnarw.
                                    .dwnarw.
                                       .dwnarw.
           M(-4)
               M(-3)
                   M(-2)
                       M(-1)
                           M(0)
                              m(1)
                                 M(2)
                                    M(3)
                                       M(4)
```

---

44 Where: M=Fourier coefficients along y-axis dimension

45 M(0)=m.sub.a (0)

46 M(1)=[m.sub.a (1)+m.sub.b (1)]/2

47 M(2)=m.sub.b (2)

48 M(3)=[m.sub.b (3)+m.sub.c (3)]/2

49 M(4)=m.sub.c (4)

50 M(-1)=[m.sub.a (-1)+m.sub.q (-1)]/2

51 M(-2)=m.sub.q (-2)

52 M(-3)=[m.sub.q (-3)+m.sub.r (-3)]/2

53 M(-4)=m.sub.r (-4)

54 The net result is that one only requires one-half of the usual number of echoes in order to fully reconstruct the image. Notice that one acquires the even projections (m(0), m(2), m(4), and so on) entirely conventionally from the signal V(-1) from coil #1. The odd projections then are calculated from both V(-1) and V(-2). The odd projections are calculated twice and averaged in the presently preferred embodiment. This provides a more artifact free reconstruction. Examples of one dimensional reconstructions are discussed below.

55 Notice that the coil to measure V(-1) is a uniform receiver coil and resembles standard designs in imaging and spectroscopy. For present purposes, this will be referred to as the "primary" channel. The coil to measure V(-2) is highly non-uniform. This will be referred to as the "secondary" channel.

56 The signal-to-noise ratio of multi (e.g., two) channel reconstruction deserves special treatment. Since, for the case N=2, one has a system of two receivers, the noise in a reconstructed image depends on whether the noise in the two channels is correlated or uncorrelated. If the coils themselves generate the noise, then the noise is uncorrelated in the two channels. Noise generated by the sample produces a noise voltage which is received by both coils and is therefore correlated. (Notice that the noise is correlated but not necessarily in phase in both channels. The relative phase of the noise voltage depends on the location of the noise generating sample.)

57 Consider a specific example. A point sample is placed at the center of the coil. In this simplified arrangement, the magnetization produced by the sample will not generate any signal in the secondary channel. So also, any noise generated by the sample will not be received by this channel. The primary channel, will receive both signal and noise generated by the sample. A quick calculation of the noise in an image with standard phase-encoded imaging and multiple receiver coil imaging shows: ##EQU13##

58 The noise at the center of the image increased by .sqroot.2. This is just a consequence of the multiple coil acquisition acquiring one-half the data. Therefore, multiple coil acquisition has the same signal-to-noise ratio per unit time as standard imaging in this example.

59 Away from the center of the image, noise actually decreases as compared to standard imaging. Since the noise is correlated in the frequency domain, the Fourier transform of the data (i.e., the image) shows structured noise.

60 This analysis changes if the noise is completely generated by the coils. Using the same example as before, the secondary channel will have a noise voltage even though there is no sample generated signal. Following in the standard noise analysis, the ratio of the noise in two imaging procedures is: ##EQU14## The noise has again increased by a factor of .sqroot.2 at the center. And, unlike the previous example, noise never decreases compared to standard imaging.

61

MRI of biological tissue in mid to high field strengths is in the regime of sample generated noise. There may be a loss in signal to noise ratio. This procedure is applicable to sequences well above the signal to noise ratio threshold where speed is the issue.

62 The expressions for the Fourier transform of the magnetization density, m(n), involve infinite sums of voltage Fourier coefficients, V(n). In practice, as proposed above, one only measures signal from a finite number of coils (e.g., two coils). The effects of truncating the infinite sum at only two terms, for example, is considered below.

63 Also, given signal from only two <u>receiver</u> coils, some Fourier projections of the magnetization density can be calculated more reliably than others. That is, the previous section contains the expressions for m(-1), m(0) and m(1) and V(-2). For the most artifact free reconstruction, calculating all three m(n)'s from one echo (resulting in a 67% time reduction), calculating two m(n)'s from one echo, or calculating three m(n)'s from one echo but redundantly calculating every other m(n) and averaging the results will provide different quality <u>MRI</u> images.

64 Considering these effects for quadrature birdcage coil <u>receivers,</u> in a typical imaging scenario, the acquired time domain signal is Fourier Transformed to yield terms in the Fourier Transform along the y direction for a given x. The remaining problem is a one-dimensional reconstruction of density along the y direction.

65 The profile of such a simple point sample is shown in FIG. 6. Here, signal only arises from a small sample located off center. A standard 128 projection reconstruction with a single <u>receiver</u> coil is shown in FIG. 7. This displays the <u>magnitude</u> of the reconstructed profile; the point spread is the usual behavior of a finitely sampled Fourier transform.

66 FIG. 8 shows the <u>magnitude</u> of a reconstruction using two <u>receivers</u> and calculating three Fourier projections per <u>echo</u>. The artifact in the reconstruction is quite large. A less ambitious approach of calculating two Fourier projections per echo (FIG. 9) with a 50% time savings decreases the artifact but it is still large.

67 The best results so far are obtained with a redundant calculation of three projections per echo; every other echo is calculated twice and averaged (as noted above in Table I). A <u>magnitude</u> reconstruction with this approach is shown in FIG. 10. The level of artifact is significantly reduced. The height of the false <u>peak</u> is roughly 2% of the height of the sample <u>peak</u>. The 50% time savings is preserved.

68 FIG. 11 shows a standard reconstruction of a point sample with sample generated noise. An image acquired in half the time using two <u>receivers</u> shows an increase in noise as shown in FIG. 12. As expected, the noise level has increased by .sqroot.2. Towards the edge of the field of view noise decreases.

69 While only a few exemplary embodiments have been described in detail above, those skilled in the art will recognize that many variations and modifications may be made in these exemplary embodiments while yet retaining many of the novel features and advantages of this invention. All such variations and modifications are intended to be encompassed by the scope of the appended claims.

CLAIMS:

What is claimed is:

1. A method for rapidly capturing MRI data, said method comprising the steps of:

(i) receiving and recording NMR RF responses in plural substantially independent RF signal receiving and processing channels during the occurrence of at least one NMR RF response; and

(ii) processing plural data sets respectively provided by said plural RF channels to produce multiply phase-encoded MRI data from said at least one NMR RF response.

2. A method as in claim 1 wherein said at least one NMR RF response is a single NMR spin echo response.

3. A magnetic resonance imaging method for materially reducing the number of repetitive NMR phase-encoded data gathering processes which must be performed to produce a multi-dimensional Fourier Transform image, said method comprising the steps of:

(i) recording NMR RF responses via a first receiving coil and RF signal processing channel to obtain first MRI data;

(ii) simultaneously recording said NMR RF responses via at least one further RF receiving coil and RF signal processing channel to obtain second MRI data;

(iii) processing said first and second MRI data to provide Fourier Transform MRI data representative of at least two different degrees of NMR phase-encoding and reducing by at least about one-half the time required for gathering NMR data sufficient to produce said multi-dimenisonal Fourier Transform image.

4. A method for magnetic resonance imaging the internal structure of an object to produce an image having resolution MxN pixels in x and y dimensions within a planar volume of substantially fixed z dimensions, MRI data for said image being captured using NMR RF responses of nuclei subjected to a static magnetic field with a sequence of mutually orthogonal x,y,z magnetic gradient pulses and NMR RF excitation pulses superimposed thereon, said method comprising the steps of:

(i) selectively exciting a planar volume of NMR nuclei of substantially fixed z dimension using a z-axis oriented magnetic gradient;

(ii) phase-encoding nuclei in a y-axis dimension using a subsequent pulse of y-axis oriented mangetic gradient;

(iii) in the presence of an x-axis oriented magnetic gradient,

(a) recording a resultant NMR RF response from a said nuclei via a first RF receiving coil and RF signal processing channel to obtain first MRI data, and

(b) simultaneously also recording said resultant NMR RF response from said nuclei via at least a second RF receiving coil and RF signal processing channel to obtain second MRI data;

(iv) processing said first and second MRI data to provide further MRI data

representative of at least two different degrees of y-axis phase encoding;

(v) repeating steps (i)-(iv) substantially less than N times, using different degrees of y-axis magnetic gradient, to provide a complete MRI data set capable of producing an MRI image having N pixel resolution along the y-axis dimension.

5. A method as in claim 4 wherein:

steps (i)-(iv) are repeated approximately N/2 times.

6. A method as in claim 4 or 5 wherein:

step (iv) provides further MRI data representative of at least three different degrees of y-axis phase encoding;

step (v) uses y-axis magnetic gradients which, on different repeat cycles, provide at least two sets of said further MRI data representing the same degree of y-axis phase encoding; and

step (v) also includes the step of averaging together said further MRI data representing the same degree of y-axis phase encoding.

7. a method as in claim 6 wherein step (iv) comprises:

(a) obtaining the first Fourier Transform coefficient V(1) of the first MRI data;

(b) obtaining the first Fourier Transform coefficient V(2) of the second MRI data;

(c) calculating three Fourier coefficients m(-1), m(0) and m(+1) of NMR magnetization for respectively corresponding different degrees of y-axis phase-encoding using the formulae:

m(-1)=iRexp(-.pi.x/R)[V(1)-i.pi.V(2)]

m(0)=iRV(1)

m(+1)=iRexp(.pi.x/R)[V(1)+i.pi.V(2)]

where

R=the effective radius of the RF receiving coils, and

x=the x coordinate for which a column of y-axis phase-encoded nuclei is being analyzed.

8. A method as in claim 7 wherein the m(-1) result for one repeat of steps (i)-(iv) is averaged with the m(+1) result for another repeat of steps (i)-(iv).

9. A magnetic resonance imaging system for imaging internal structure of an

object using NMR RF responses of nuclei subjected to a static magnetic field and having a sequence of magnetic gradients and RF NMR excitation pulses superimposed thereon, said system comprising:

a first RF coil coupled to a predetermined image volume for receiving said NMR RF responses with a first relationship between the spatial location of NMR nuclei and a resultive first RF signal output from said first RF coil;

a second RF coil also coupled to said image volume, for receiving said NMR RF responses with a second relationship between the spatial location of NMR nuclei and a resulting second RF signal output from said second RF coil;

said first and second RF coils being substantially independent and uncoupled to one another insofar as the NMR RF responses are concerned;

first and second independent RF signal processing channels connected respectively to individually receive and process said first and second RF signal outputs to produce independent first and second NMR response data from a single NMR RF response event; and

image processing means connected to receive said independent first and second NMR response data and to produce multiply phase-encoded NMR image data from said single NMR RF response event.

10. A magnetic resonance imaging system as in claim 9 wherein:

said first and second RF coils each comprise birdcage coils, said first RF coil being resonant at a first fundamental frequency of about F and said second RF coil being resonant at a second fundamental frequency of about F.

11. A magnetic resonance imaging system as in claim 9 wherein said first and second RF coils each comprise a pair of QD-connected saddle coils, said first RF coil being resonant at a first fundamental frequency of about F and said second RF coil being resonant at a second fundamental frequency of about F.

12. A magnetic resonance imaging system as in claim 9, 10, or 11 wherein:

said first RF coil provides a current/voltage distribution which varies as sin .theta. about said image volume; and

said second RF coil provides a current/voltage distribution which varies as sin 2.theta. about said image volume.

13. Apparatus for rapidly capturing MRI data, said apparatus comprising:

(i) means for receiving and recording NMR RF responses in plural substantially independent RF signal receiving and processing channels during the occurrence of at least one NMR RF response; and

(ii) means for processing plural data sets respectively provided by said plural RF channels to produce multiply phase-encoded MRI data from said at least one NMR RF response.

14. Magnetic resonance imaging apparatus for materially reducing the number of repetitive NMR phase-encoded data gathering processes which must be performed to produce a multi-dimensional Fourier Transform image, said apparatus

comprising: of:

(i) means for recording NMR RF responses via a first receiving coil and RF signal processing channel to obtain first MRI data;

(ii) means for simultaneously recording said NMR RF responses via at least one further RF receiving coil and RF signal processing channel to obtain second MRI data;

(iii) means for processing said first and second MRI data to provide Fourier Transform MRI data representative of at least two different degrees of NMR phase-encoding and reducing by at least about one-half the time required for gathering NMR data sufficient to produce said multi-dimenisonal Fourier Transform image.

15. A apparatus for magnetic resonance imaging the internal structure of an object to produce an image having resolution MxN pixels in x and y dimensions within a planar volume of substantially fixed z dimensions, MRI data for said image being captured using NMR RF responses of nuclei subjected to a static magnetic field with a sequence of mutually orthogonal x,y,z magnetic gradient pulses and NMR RF excitation pulses superimposed thereon, said apparatus comprising:

(i) means for selectively exciting a planar volume of NMR nuclei of substantially fixed z dimension using a z-axis oriented magnetic gradient;

(ii) means for phase-encoding nuclei in a y-axis dimension using a subsequent pulse of y-axis oriented mangetic gradient;

(iii) means for, in the presence of an x-axis oriented magnetic gradient,

(a) recording a resultant NMR RF response from a said nuclei via a first RF receiving coil and RF signal processing channel to obtain first MRI data, and

(b) simultaneously also recording said resultant NMR RF response from said nuclei via at least a second RF receiving coil and RF signal processing channel to obtain second MRI data;

(iv) means for processing said first and second MRI data to provide further MRI data representative of at least two different degrees of y-axis phase encoding;

(v) means for repeating said exciting, phase-encoding and recording substantially less than N times, using different degrees of y-axis magnetic gradient, to provide a complete MRI data set capable of producing an MRI image having N pixel resolution along the y-axis dimension.

16. Apparatus as in claim 15 wherein:

said means for repeating causes approximately N/2 repetitions to occur.

17. A apparatus as in claim 15 or 16 wherein:

said means processing provides further MRI data representative of at least three different degrees of y-axis phase encoding;

said means for repeating uses y-axis magnetic gradients which, on different repeat cycles, provide at least two sets of said further MRI data representing the same degree of y-axis phase encoding; and

said means for repeating also includes means for averaging together said further MRI data representing the same degree of y-axis phase encoding.

18. Apparatus as in claim 17 wherein said means for processing comprises:

(a) means for obtaining the first Fourier Transform coefficient V(1) of the first MRI data;

(b) means for obtaining the first Fourier Transform coefficient V(2) of the second MRI data;

(c) means for calculating three Fourier coefficients m(-1), m(0) and m(+1) of NMR magnetization for respectively corresponding different degrees of y-axis phase-encoding using the formulae:

m(-1)=iRexp(-.pi.x/R)[V(1)-i.pi.V(2)]

m(0)=iRV(1)

m(+1)=iRexp(.pi.x/R)[V(1)+i.pi.V(2)]

where

R=the effective radius of the RF receiving coils, and

x=the x coordinate for which a column of y-axis phase-encoded nuclei is being analyzed.

19. Apparatus as in claim 18 wherein said means for averaging averages the m(-1) result for one repeat with the m(+1) result for another repeat.

20. A magnetic resonance imaging method for imaging internal structure of an object using NMR RF responses of nuclei subjected to a static magnetic field and having a sequence of magnetic gradients and RF NMR excitation pulses superimposed thereon, said method comprising:

receiving said NMR RF responses with a first relationship between the spatial location of NMR nuclei and a resultive first RF signal output from a first RF coil;

receiving said NMR RF responses with a second relationship between the spatial location of NMR nuclei and a resulting second RF signal output from a second RF coil;

said first and second RF coils being substantially independent and uncoupled to one another insofar as the NMR RF responses are concerned;

individually receiving and processing said first and second RF signal outputs to produce independent first and second NMR response data from a single NMR RF response event; and

receiving said independent first and second NMR response data and producing multiply phase-encoded NMR image data from said single NMR RF response event.

21. A magnetic resonance imaging method as in claim 20 wherein:

said first and second RF coils used in said receiving NMR RF responses steps respectively each comprise birdcage coils, said first RF coil being resonant at a first fundamental frequency of about F and said second RF coil being resonant at a second fundamental frequency of about F.

22. A magnetic resonance imaging method as in claim 20 wherein said first and second RF coils used in said receiving NMR RF response steps respectively each comprise a pair of QD-connected saddle coils, said first RF coil being resonant at a first fundamental frequency of about F and said second RF coil being resonant at a second fundamental frequency of about F.

23. A magnetic resonance imaging method as in claim 20, 21 or 22 wherein:

said first RF coil provides a current/voltage distribution which varies as sin .theta. about said image volume; and

said second RF coil provides a current/voltage distribution which varies as sin 2.theta. about said image volume.

Previous Doc Next Doc Go to Doc#

Hit List

Clear | Generate Collection | Print | Fwd Refs | Bkwd Refs
Generate OACS

Search Results - Record(s) 1 through 9 of 9 returned.

☐ 1. Document ID: US 20040155652 A1

Using default format because multiple data bases are involved.

L19: Entry 1 of 9 File: PGPB Aug 12, 2004

PGPUB-DOCUMENT-NUMBER: 20040155652
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20040155652 A1

TITLE: Parallel magnetic resonance imaging techniques using radiofrequency coil arrays

PUBLICATION-DATE: August 12, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Sodickson, Daniel K.	Newton	MA	US	

US-CL-CURRENT: 324/307; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw De

---

☐ 2. Document ID: US 20040044280 A1

L19: Entry 2 of 9 File: PGPB Mar 4, 2004

PGPUB-DOCUMENT-NUMBER: 20040044280
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20040044280 A1

TITLE: Methods & apparatus for magnetic resonance imaging

PUBLICATION-DATE: March 4, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Paley, Martyn	Keighly		GB	
Lee, Kuan	Sheffield		GB	

US-CL-CURRENT: 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw De

☐ 3. Document ID: US 20030206648 A1

L19: Entry 3 of 9 File: PGPB Nov 6, 2003

PGPUB-DOCUMENT-NUMBER: 20030206648
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20030206648 A1

TITLE: Method and system for image reconstruction

PUBLICATION-DATE: November 6, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
King, Kevin Franklin	New Berlin	WI	US	
Angelos, Elisabeth	Hartland	WI	US	

US-CL-CURRENT: 382/128

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw D

☐ 4. Document ID: US 20020158632 A1

L19: Entry 4 of 9 File: PGPB Oct 31, 2002

PGPUB-DOCUMENT-NUMBER: 20020158632
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020158632 A1

TITLE: Parallel magnetic resonance imaging techniques using radiofrequency coil arrays

PUBLICATION-DATE: October 31, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Sodickson MD Ph.D., Daniel K.	Cambridge	MA	US	

US-CL-CURRENT: 324/307; 324/309, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw D

☐ 5. Document ID: US 20010043068 A1

L19: Entry 5 of 9 File: PGPB Nov 22, 2001

PGPUB-DOCUMENT-NUMBER: 20010043068
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20010043068 A1

TITLE: Method for parallel spatial encoded MRI and apparatus, systems and other methods related thereto

PUBLICATION-DATE: November 22, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Lee, Ray F.	Clifton-Park	NY	US	

US-CL-CURRENT: 324/309; 324/307, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw De

---

☐ 6. Document ID: US 6841998 B1

L19: Entry 6 of 9 File: USPT Jan 11, 2005

US-PAT-NO: 6841998
DOCUMENT-IDENTIFIER: US 6841998 B1

TITLE: Magnetic resonance imaging method and apparatus employing partial parallel acquisition, wherein each coil produces a complete k-space datasheet

DATE-ISSUED: January 11, 2005

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Griswold; Mark	97318 Kitzingen			DE

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw De

---

☐ 7. Document ID: US 6771067 B2

L19: Entry 7 of 9 File: USPT Aug 3, 2004

US-PAT-NO: 6771067
DOCUMENT-IDENTIFIER: US 6771067 B2

TITLE: Ghost artifact cancellation using phased array processing

DATE-ISSUED: August 3, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Kellman; Peter	Bethesda	MD		
McVeigh; Elliot	Phoenix	MD		

US-CL-CURRENT: 324/307; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw. D

☐ 8. Document ID: US 6717406 B2

L19: Entry 8 of 9 File: USPT Apr 6, 2004

US-PAT-NO: 6717406
DOCUMENT-IDENTIFIER: US 6717406 B2

TITLE: Parallel magnetic resonance imaging techniques using radiofrequency coil arrays

DATE-ISSUED: April 6, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Sodickson; Daniel K.	Newton	MA		

US-CL-CURRENT: 324/307; 324/309, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw. D

☐ 9. Document ID: US 6476606 B2

L19: Entry 9 of 9 File: USPT Nov 5, 2002

US-PAT-NO: 6476606
DOCUMENT-IDENTIFIER: US 6476606 B2

TITLE: Method for parallel spatial encoded MRI and apparatus, systems and other methods related thereto

DATE-ISSUED: November 5, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Lee; Ray F	Clifton-Park	NY		

US-CL-CURRENT: 324/309; 324/307, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw. D

Clear | Generate Collection | Print | Fwd Refs | Bkwd Refs | Generate OACS

Term	Documents
MULTI	1047247

MULTIS	135
COIL	1219317
COILS	404096
IMAGE	2517743
IMAGES	589159
COMBIN$3	0
COMBIN	30172
COMBINA	80369
COMBINAA	154
COMBINAAA	1
(L18 AND ((COMBIN$3 OR (MULTI WITH COIL)) WITH (IMAGE OR RECONSTRUCT$4))).PGPB,USPT,USOC,EPAB,JPAB,DWPI,TDBD.	9

There are more results than shown above. Click here to view the entire set.

Display Format: - Change Format

Previous Page Next Page Go to Doc#

Hit List

Clear | Generate Collection | Print | Fwd Refs | Bkwd Refs | Generate OACS

Search Results - Record(s) 1 through 12 of 12 returned.

☐ 1. Document ID: US 20050033154 A1

Using default format because multiple data bases are involved.

L20: Entry 1 of 12 File: PGPB Feb 10, 2005

PGPUB-DOCUMENT-NUMBER: 20050033154
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20050033154 A1

TITLE: Methods for measurement of magnetic resonance signal perturbations

PUBLICATION-DATE: February 10, 2005

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
deCharms, Richard Christopher	Montara	CA	US	

US-CL-CURRENT: 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw D

☐ 2. Document ID: US 20040155652 A1

L20: Entry 2 of 12 File: PGPB Aug 12, 2004

PGPUB-DOCUMENT-NUMBER: 20040155652
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20040155652 A1

TITLE: Parallel magnetic resonance imaging techniques using radiofrequency coil arrays

PUBLICATION-DATE: August 12, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Sodickson, Daniel K.	Newton	MA	US	

US-CL-CURRENT: 324/307; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw D

☐ 3. Document ID: US 20040044280 A1

L20: Entry 3 of 12 File: PGPB Mar 4, 2004

PGPUB-DOCUMENT-NUMBER: 20040044280
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20040044280 A1

TITLE: Methods & apparatus for magnetic resonance imaging

PUBLICATION-DATE: March 4, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Paley, Martyn	Keighly		GB	
Lee, Kuan	Sheffield		GB	

US-CL-CURRENT: 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw De

☐ 4. Document ID: US 20030206648 A1

L20: Entry 4 of 12 File: PGPB Nov 6, 2003

PGPUB-DOCUMENT-NUMBER: 20030206648
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20030206648 A1

TITLE: Method and system for image reconstruction

PUBLICATION-DATE: November 6, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
King, Kevin Franklin	New Berlin	WI	US	
Angelos, Elisabeth	Hartland	WI	US	

US-CL-CURRENT: 382/128

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw De

☐ 5. Document ID: US 20020158632 A1

L20: Entry 5 of 12 File: PGPB Oct 31, 2002

PGPUB-DOCUMENT-NUMBER: 20020158632
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020158632 A1

TITLE: Parallel magnetic resonance imaging techniques using radiofrequency coil arrays

PUBLICATION-DATE: October 31, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Sodickson MD Ph.D., Daniel K.	Cambridge	MA	US	

US-CL-CURRENT: 324/307; 324/309, 324/318

☐ 6. Document ID: US 20010043068 A1

L20: Entry 6 of 12 File: PGPB Nov 22, 2001

PGPUB-DOCUMENT-NUMBER: 20010043068
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20010043068 A1

TITLE: Method for parallel spatial encoded MRI and apparatus, systems and other methods related thereto

PUBLICATION-DATE: November 22, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Lee, Ray F.	Clifton-Park	NY	US	

US-CL-CURRENT: 324/309; 324/307, 324/318

☐ 7. Document ID: US 6841998 B1

L20: Entry 7 of 12 File: USPT Jan 11, 2005

US-PAT-NO: 6841998
DOCUMENT-IDENTIFIER: US 6841998 B1

TITLE: Magnetic resonance imaging method and apparatus employing partial parallel acquisition, wherein each coil produces a complete k-space datasheet

DATE-ISSUED: January 11, 2005

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Griswold; Mark	97318 Kitzingen			DE

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Drawn D

☐ 8. Document ID: US 6771067 B2

L20: Entry 8 of 12 File: USPT Aug 3, 2004

US-PAT-NO: 6771067
DOCUMENT-IDENTIFIER: US 6771067 B2

TITLE: Ghost artifact cancellation using phased array processing

DATE-ISSUED: August 3, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Kellman; Peter	Bethesda	MD		
McVeigh; Elliot	Phoenix	MD		

US-CL-CURRENT: 324/307; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Drawn D

☐ 9. Document ID: US 6717406 B2

L20: Entry 9 of 12 File: USPT Apr 6, 2004

US-PAT-NO: 6717406
DOCUMENT-IDENTIFIER: US 6717406 B2

TITLE: Parallel magnetic resonance imaging techniques using radiofrequency coil arrays

DATE-ISSUED: April 6, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Sodickson; Daniel K.	Newton	MA		

US-CL-CURRENT: 324/307; 324/309, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Drawn D

☐ 10. Document ID: US 6476606 B2

L20: Entry 10 of 12 File: USPT Nov 5, 2002

US-PAT-NO: 6476606
DOCUMENT-IDENTIFIER: US 6476606 B2

TITLE: Method for parallel spatial encoded MRI and apparatus, systems and other methods related thereto

DATE-ISSUED: November 5, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Lee; Ray F	Clifton-Park	NY		

US-CL-CURRENT: 324/309; 324/307, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☑ 11. Document ID: US 4999581 A

L20: Entry 11 of 12 File: USPT Mar 12, 1991

US-PAT-NO: 4999581
DOCUMENT-IDENTIFIER: US 4999581 A

TITLE: Magnetic resonance imaging system

DATE-ISSUED: March 12, 1991

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Satoh; Kozo	Yokohama			JP

US-CL-CURRENT: 324/309; 324/314

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☑ 12. Document ID: US 4857846 A

L20: Entry 12 of 12 File: USPT Aug 15, 1989

US-PAT-NO: 4857846
DOCUMENT-IDENTIFIER: US 4857846 A

TITLE: Rapid MRI using multiple receivers producing multiply phase-encoded data derived from a single NMR response

DATE-ISSUED: August 15, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Carlson; Joseph W.	San Francisco	CA		

US-CL-CURRENT: 324/309; 324/314

Clear | Generate Collection | Print | Fwd Refs | Bkwd Refs | Generate OACS

Term	Documents
MULTI	1047247
MULTIS	135
COIL	1219317
COILS	404096
IMAGE	2517743
IMAGES	589159
COMBIN$3	0
COMBIN	30172
COMBINA	80369
COMBINAA	154
COMBINAAA	1
(L18 AND ((COMBIN$3 OR "MULTI" OR COIL) WITH (IMAGE OR RECONSTRUCT$4))).PGPB,USPT,USOC,EPAB,JPAB,DWPI,TDBD.	12

There are more results than shown above. Click here to view the entire set.

Display Format: - Change Format

Previous Page Next Page Go to Doc#

L21: Entry 50 of 54 File: DWPI Apr 20, 1995

DERWENT-ACC-NO: 1995-156459
DERWENT-WEEK: 199521

TITLE: Medical MRI arrangement with dynamic receiver gain - uses normalisation table to determine amplitude and phase correction values for MR signals acquired via receivers with gains set using associated damping values

INVENTOR: MAIER, J K; SANDFORD, L V ; STORMONT, R S

PATENT-ASSIGNEE: GENERAL ELECTRIC CO (GENE)

PRIORITY-DATA: 1993US-0138273 (October 18, 1993)

Search Selected Search ALL Clear

PATENT-FAMILY:

PUB-NO	PUB-DATE	LANGUAGE	PAGES	MAIN-IPC
DE 4436801 A1	April 20, 1995		009	G01N024/08
US 5451876 A	September 19, 1995		008	G01V003/00

APPLICATION-DATA:

PUB-NO	APPL-DATE	APPL-NO	DESCRIPTOR
DE 4436801A1	October 14, 1994	1994DE-4436801	
US 5451876A	October 18, 1993	1993US-0138273	

INT-CL (IPC): A61 B 5/055; G01 N 24/08; G01 V 3/00

ABSTRACTED-PUB-NO: DE 4436801A
BASIC-ABSTRACT:

A normalisation table is stored which contains several amplitude and phase correction values. Each amplitude and phase correction value is associated with a corresp. one of a set of receiver damping values. A MR signal is acquired via a receiver whose gain is set using one of the receiver damping values.

The acquired MR signal is normalised by varying its amplitude and phase by an amount determined from the amplitude and phase correction values in the normalisation table associated with the receiver damping value. The process is repeated to acquire further MR signals with receiver gains set using other receiver damping values. The image is reconstructed using the normalised acquired MR signals.

USE/ADVANTAGE - Regeneration of image from several derived MR signals for human body investigation. Improved MRI arrangement enables improved noise separation of magnetic resonance images to be achieved.

ABSTRACTED-PUB-NO: US 5451876A
EQUIVALENT-ABSTRACTS:

In an NMR system a method for reconstructing an image from a plurality of acquired NMR signals, the steps comprising:

a) storing a normalization table comprised of a plurality of amplitude and phase corrections (A,theta), each of the amplitude and phase corrections being associated with one of a corresponding plurality of receiver attenuation values (RA);

b) acquiring one of said NMR signals with a receiver whose gain is set by one of said receiver attenuation values (RA);

c) normalizing the acquired NMR signal by altering its amplitude and phase by an amount determined by the amplitude and phase corrections (A,theta) in the normalization table associated with said one of said receiver attenuation values (RA);

d) repeating steps b) and c) to acquire further NMR signals with the receiver gain set by different ones of said receiver attenuation values (RA); and

e) reconstructing an image using the normalized, acquired NMR signals.

CHOSEN-DRAWING: Dwg.1/4 Dwg.2/4

DERWENT-CLASS: P31 S03 S05 T01
EPI-CODES: S03-E07A; S05-D02B2; T01-J10A;

Previous Doc Next Doc Go to Doc#